

## DOCTORAL THESIS

### **An initial investigation into the effect of pain relief on lumbar kinematics and electromyography in low back pain sufferers**

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**An initial investigation into the effect of pain  
relief on lumbar kinematics and  
electromyography in low back pain sufferers**

**by**

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**A thesis submitted in partial fulfilment of the requirements  
for the degree of PhD**

**Department of Life Sciences**

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## **Abstract**

Motion and motor patterns of low back pain (LBP) sufferers have been demonstrated as different compared to those without LBP. The mechanism behind such movement alterations is not well known but is believed to be related to pain. Current biomechanical measurement of lumbar curvature and kinematics has limitations for routine clinical use. The aims of this research were to (1) investigate new motion analysis technology for dynamic lumbar curvature (fibre-optic sensors) and higher order kinematic assessment (inertial sensors) within a clinical environment; (2) determine the effect of pain relief on lumbar curvature, kinematics and muscle function in acute low back pain (ALBP) and chronic low back pain (CLBP) sufferers. Dynamic lumbar curvature was found to be reliably measureable in the clinic. Additional analysis demonstrated that regional curvature, as well as sequencing of curvature change measurement was possible. ALBP sufferers display less peak curvature during flexion and lifting compared to CLBP sufferers and both groups demonstrated the greatest curvature change in the second quartile for flexion and lifting and first for extension. Partial pain relief did not increase curvature in either group, and neither group was more likely to respond to pain relief by increasing curvature or altering sequencing. Higher order kinematics could be reliably and readily identified in the clinic using inertial sensors. Movement-velocity plots were employed to describe the movement trajectory and irregularity. Multivariate analysis of variance revealed that neither partial pain relief (pre vs post) nor chronicity (acute vs chronic) had any effect on lumbar kinematics. Individuals appeared to exhibit different electromyography profiles. Those individuals with little deviation in muscle activation commonly displayed lower kinematic values. Partial pain relief did not alter the pattern of EMG profile, muscle onset timing or peak amplitude. These results suggest that clinicians should not expect automatic alterations in motion and motor patterns following interventions which target and achieve partial pain relief.

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## **Publications**

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Williams, JM, Haq, I, Lee, RY. 2010. Dynamic measurement of lumbar curvature using fibre-optic sensors. *Medical Engineering and Physics*. 32 (9): 1043-1049.

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# **1 Introduction**

## **1.1 Statement of the problem**

Low Back Pain (LBP) is a widespread problem with nearly everyone affected at some time in their lives (Savigny et al., 2009). Prevalence estimates range from 12-33% for point prevalence, 22-65% for 1-year prevalence and up to 84% for lifetime prevalence (Walker, 2000).

Prevalence estimates vary due to different definitions of what constitutes LBP (Dionne et al., 2008). It has been shown that one year following an initial episode of LBP, between 42-75% of individuals were still suffering from LBP and 3-40% remained sick-listed at one year follow up (Hestbaek, Leboeuf-Yde, & Manniche, 2003). The direct healthcare costs of LBP for the UK have been estimated at £1632 million, but only as recently as 1998 (Maniadakis & Gray, 2000). Indirect costs such as lost productivity were much higher at £9090 million. The direct costs of physiotherapy alone were \$150.7 million and £100.5 million for NHS and private physiotherapy respectively (Maniadakis et al., 2000).

The assessment and management of LBP often involves the observation of motion and motor behaviours which are then used as a framework for therapeutic intervention. Indeed altered motion and motor patterns are frequently observed in LBP sufferers (Geisser et al., 2005; Marras et al., 1999; Marras et al., 1995). Kinematics and muscle behaviours are known to largely influence spinal loading and are linked to failure of spinal structures (Adams & Hutton, 1982; Adams, Hutton, & Stott, 1980b; Adams, May, Freeman, Morrison, & Dolan, 2000; Dolan et al., 1999; Gallagher, Marras, Litsky, & Burr, 2005). Furthermore they are noted as risk factors for LBP development and LBP reporting (Magora, 1973; Norman et al., 1998). It is known that persons with LBP are likely to display altered kinematics and altered trunk muscle function when compared with matched controls and that the resolution of pre-pain biomechanical functions is not guaranteed on cessation of the LBP episode (Ferguson, Marras, & Gupta, 2000; Hodges, van den Hoorn, Dawson, & Cholewicki, 2009; McGill et al., 2003). It is

thought that this may offer a mechanical mechanism for ongoing spinal pain and continued disability. Indeed even long after the recovery from the LBP episode, deficits in function of the low back remain evident (McGill et al., 2003). It is important therefore to be able to objectively measure spinal functions within the clinic and monitor recovery of biomechanical functions of the spine. In order to achieve this aim it is imperative that clinic based motion analysis systems are developed which are able to measure important kinematic variables within the constraints of the clinical environment.

However just observing movement and motor behaviour alterations is insufficient in truly understanding why these individuals move the way they do. Management strategies often involve either targeting pain relief to alter spinal biomechanical functions (Jette, Smith, Haley, & Davis, 1994) or directly aim to influence movements and motor behaviour (O'Sullivan, 2005). If the aim of treatment is to restore the biomechanical functions of the spine then it is imperative to identify the underlying cause or mechanism responsible for the biomechanical changes so that clinical management can be rationalised. Unfortunately current knowledge regarding this is limited.

## **1.2 Purpose of the study**

The purpose of the first section of the study was to evaluate spinal curvature using a novel fibre-optic motion analyses system, within a clinical setting, to overcome limitations of currently available curvature measurement methods. Information from a string of sensors attached to the skin overlying the spine provides a measure of overall spinal shape, dynamically across a range of functional tasks. Curvature measurements were calculated for a series of known points along the spine corresponding to the location of specific spinous processes. The reliability and validity of such a system was initially tested by completing a comparative study with an opto-electronic video based laboratory system (Vicon). The system

was then utilised to investigate the curvature behaviour in acute and chronic LBP sufferers and to investigate the effects of pain-relief on lumbar curvature.

The thesis continues by investigating the use of inertial sensor technology to measure higher order kinematics and extends the analysis to include the spatial domain. The feasibility of using this spatial domain analysis to provide a measure of movement irregularity is presented and the thesis proceeds to investigate the effects of pain-relief on lumbar kinematics including the higher order kinematics and movement irregularity, during spinal movements in acute and chronic LBP sufferers.

The purpose of the third section of the thesis was to investigate the effects of pain relief on the muscle functions of iliocostalis and multifidus. The pattern of EMG profile was classified and cross-correlated to the movement, with pattern, muscle onset time and peak compared before and after analgesia to determine whether pain relief alters muscle function of these paraspinal muscles.

This thesis presents results from the studies outlined above. It is hoped that the application of new technology may offer a solution to the problem of dynamic curvature measurement in the spine. Additionally information about varying regions is collected simultaneously enabling the option of regional and possibly segmental analysis of the spine during movement. The extension of analysis using a clinic based inertial sensor system provides clinicians with the ability to gain a more thorough kinematic profile including insights into movement irregularity through the spatial domain. The knowledge gained from the determination of the effects of pain-relief, on lumbar curvature, kinematics and muscle functions, enables the development of our understanding as to the role of pain in driving or maintaining movement behaviour change. This knowledge will be invaluable in the development of clinical assessment and management strategies.

### 1.3 Organisation of Thesis

Chapter 1 provides an introduction and outlines the purpose and organisation of the thesis.

Chapter 2 reviews the related literature pertaining to motion analysis and curvature measurement methods and technologies for the lumbar spine, in both a clinical and laboratory setting. It continues with a contemporary review of the literature pertaining to the question is pain the cause of altered biomechanical functions in back pain sufferers.

Chapter 3 presents the studies related to a fibre-optic motion analysis system. The first section presents data pertaining to the reliability and validity of the fibre-optic system. The second section investigates curvature in LBP sufferers including peak curvature and lumbar sequencing of curvature change. The final section determines the effects of pain relief on curvature and sequencing in LBP sufferers.

Chapter 4 presents studies pertaining to the use of inertial sensors in measuring lumbar kinematics. The first section demonstrates the repeated measures reliability of higher order kinematics and investigates the use of spatial plots for visualisation of the movement trajectory to provide an insight into movement coordination and irregularity. The second section investigates the effects of pain relief on lumbar kinematics.

Chapter 5 presents results for the effects of pain relief on spinal muscle function (multifidus and iliocostalis). The chapter investigates the effects of pain relief on EMG profile, onset timing and peak activation.

Chapter 6 presents a general discussion of findings relating to this thesis.

Chapter 7 presents conclusions and recommendations for future work.

## **Chapter 2:**

### **Review of related literature**

## **2 Review of related literature**

A literature review of currently available methods for measuring spinal kinematics (section 2.1) is presented. This chapter evaluates methods broadly classifying them as either clinic based or laboratory based systems. Clinical methods are often quick and simple but have limitations in the amount of kinematic information available. Laboratory systems are often expensive and have environmental constraints as well as involving complex data processing techniques. The review focuses on reliability and validity but proceeds also to discuss advantages and disadvantages of each system including limitations and potential solutions. The review then continues by critiquing the available methods for the measurement of lumbar curvature covering the advantages and limitations of each method presented (section 2.1.3). This section concludes with a summary and potential suggested future directions. The final section of the review will explore the literature underpinning whether pain is the mechanisms behind alterations in biomechanical behaviour of the spine (section 2.2).

### **2.1 Measurement techniques for lumbar kinematics**

Measurement of lumbar spine curvature and motion has become common place in the clinical assessment of low back pain (LBP). It provides useful information regarding spinal function and is often used as an outcome measure following clinical interventions (Cocchiarella & Andersson, 2000; Magnusson et al., 1998). Due to its unique anatomy and complex movements the measurement of curvature and motion of the lumbar spine is technically challenging. Various measurement options are available ranging from simple visual estimation and inclinometers, within the clinic, to complex 3-dimensional video or electromagnetic methods found mainly in research laboratories. Despite this clinicians and researchers require the chosen method to be both reliable and valid, therefore the aim of this review is to compare the reliability and validity, as well as contrast and critique various lumbar spine measurement methods for resolving curvature and motion of the lumbar spine.

### **2.1.1 Clinic based measurement tools**

#### **2.1.1.1 Skin Distraction**

Skin distraction techniques involve marking and measuring the distance between two points on the skin usually overlying a bony landmark (spinous process) in different functional positions. It represents a quick, simple, cheap and non-invasive method of measuring lumbar spinal motion. Use of skin distraction techniques was originally described by Schober (Schober, 1937) and was shown to have a correlation coefficient of 0.90 when compared with uniplaner radiography (MacRae & Wright, 1969). Furthermore, a small modification of the original technique improved the correlation coefficient to 0.97 (MacRae et al., 1969). However more recently it has been suggested that skin distraction is influenced more by body height and weight than spinal movement (Dolan, Mannion, & Adams, 1995). This may suggest the degree of skin deformation is affected by anthropometric factors confounding the results (Miller, Mayer, Cox, & Gatchel, 1992). Further limitations of the skin-distraction method include the inability to measure movements which are not sagittal or to gain information about out of plane movements. Furthermore no knowledge of the time history or pattern of the movement is possible as it provides end-point static measurement only.

In summary therefore, despite offering a quick and simple measurement method of spinal flexion with good correlation to radiographs, however the limitations mean it cannot provide information regarding the time related movement behaviour of the lumbar spine.

#### **2.1.1.2 Inclinator**

An inclinometer is a device that measures the angle of inclination relative to a reference i.e. the vertical. It offers an attractive option for spinal posture and motion analysis as it quick and simple and can measure various spinal regions, as well as being able to differentiate between pelvic and lumbar contributions to overall motion (Ng, Kippers, Richardson, & Parnianpour,

2001). Intra-rater reliability for flexion have been shown as good with intraclass correlation coefficients of 0.79 – 0.91 (Burdett, Brown, & Fall, 1986; Kachingwe & Phillips, 2005; Ng et al., 2001). Measurements of extension however are less reliable, ranging from 0.60 -0.75 (Kachingwe et al., 2005; Saur, Ensink, Frese, Seeger, & Hildebrandt, 1996) , with only one study reporting excellent reliability (Ng et al., 2001). Reliability of side bending measurements has been shown to range between 0.83-0.92 (Kachingwe et al., 2005; Ng et al., 2001).

Inter-rater reliability has shown large inconsistencies, with extension displaying the greatest discrepancies across operators with correlation coefficients ranging from 0.42-0.55 (Kachingwe et al., 2005; Saur et al., 1996) whilst for flexion it is reported as 0.74-0.88 (Kachingwe et al., 2005; Saur et al., 1996) . This may represent small disagreements in methods of data collection between operators, for example different spinous process location techniques or varied pressure applied when using the device.

Validity of inclinometers has been tested using direct comparison with radiographic measurements and has suggested poor correlation to actual spinal movements. Saur et al., (1996) reported promising results with pearson correlation coefficients of  $r = 0.80$  and  $0.75$  for flexion and extension respectively. However, both Bierma-Zeinstra et al., (2001) and Burdett et al., (1986) provide data to the contrary. These authors report good validity for flexion but poor validity for extension ( $r = 0.73$  and  $0.15$ ) (Burdett et al., 1986), along with  $r = 0.28$  (Bierma-Zeinstra et al., 2001) for standing. Furthermore this translated to a mean difference between radiographic and inclinometer measurement to be  $23.12 \pm 8.56^\circ$ .

Further limitations of this method are in its inability to measure the time-history of movement as the method requires that a static posture must be adopted in order to place and read the inclinometer. Also it is unable to provide information regarding axial rotation, as it uses the



pendulum principle, and is unable to provide information about movements out of the plane of interest.

In summary it seems that the inclinometer offers a method of motion analysis that can determine lumbar motion from pelvic motion, with good intra-rater reliability. Inter-rater reliability is more questionable. It is likely that the results obtained represent a mobility index rather than direct representation of actual true spinal motion due to poor correlations with radiographs. This is possible in the sagittal and frontal plane, however extra caution should be applied when interpreting extension due to the large measurement error associated with this motion. However inclinometers cannot resolve axial rotation, dynamic motion or motion out of the plane of interest and therefore provide only limited information about spinal movement behaviour.

#### **2.1.1.3 Fingertip-to-floor**

Measurement of lumbar motion using the distance from the finger-tips to the floor has been utilised as way of measuring flexion, extension and side bending motion in the clinical setting. It is very quick and simple to employ. Reliability measures have shown promising results for flexion, with intra-rater correlation coefficients ranging from 0.95-0.98 and inter-rater correlation coefficients of 0.98 (Frost, Stuckey, Smalley, & Dorman, 1982; Gauvin, Riddle, & Rothstein, 1990). It is also reported that side-bending measurements display good inter- and intra-rater reliability ( $r = 0.91$ ) (Frost et al., 1982). Backward bending results in values slightly less of 0.78 and 0.79 for inter- and intra-rater reliability respectively (Frost et al., 1982).

The validity of such methods to measure lumbar motion has been questioned due to the varying contributions of thoracic, pelvic and limb flexibility to the overall result (Battie, Bigos, Sheehy, & Wortley, 1987; Porter & Wilkinson, 1997; Wong & Lee, 2004). Further limitations of

such a method are that it is only able to measure static positions and is limited to one plane therefore no information regarding dynamic movement behaviour is possible.

In summary the finger-tip-to-floor method offers a quick, cheap and reliable method of assessing spinal motion, however its underlying construct validity is questionable due to varying contributions from other anatomical locations. Understanding whether a change in result is due to a change in lumbar movement is therefore not possible. Further limitations make its use difficult for understanding the time related movement behaviour of the spine.

#### **2.1.1.4 Flexicurve**

Flexicurve measurement utilises a modified draftsman's curve which is moulded over the spine and traced to paper, where pen and paper calculations are completed to measure curvature. Standing curvatures are then subtracted from end range curvatures to yield ROM values (Burton, 1987; Burton & Tillotson, 1989; Burton, 1986). It is cheap and simple to use, however measurement is confined to the single plane of interest and can only be used to measure static postures.

Intra-observer reliability has been reported at between 0.95-0.97 for flexion and extension movements, resulting in a maximum technical error of measurement of  $1.4^{\circ}$ . Inter-rater reliability ranged from 0.82-0.99 with maximal technical error of  $2.5^{\circ}$  (Burton, 1986).

Validity has been tested by direct comparison to lateral radiographs, suggesting an agreement to within  $1^{\circ}$ , however only one subject was tested (Burton, 1986).

This method may offer important information not readily available from other clinic based systems. The flexicurve is able to provide a continuous representation of spinal shape therefore providing the along-spine distances are known curvature and range of motion for

any 'region' can be determined, as demonstrated by measuring upper and lower lumbar regions (Burton, 1987; Burton, 1986). Unfortunately this is only possible for static postures and in one plane, offering no information regarding the time-history of motion.

#### **2.1.1.5 Electrogoniometer**

The electrogoniometer is made up of two end plates with strain gauge wire or potentiometer connected between. It operates on the principle that as the individual bends this is sensed in the wire or potentiometer, therefore providing a measure of the angle between the two end plates attached to the skin (Rowe, Myles, Hillmann, & Hazlewood, 2001). It offers significant advantage over other clinical systems as it can provide dynamic measurement of angle, enabling the dynamic behaviour of a joint to be investigated. Furthermore it can provide data for up to two planes of motion simultaneously.

The reliability and accuracy has been tested using jig based experiments where repeated measures have been found to be within  $0.1^{\circ}$  over a whole range of angles (Christensen, 1999). The comparative accuracy to precision protractors was  $< 11.5\%$  for all movements, however clearly the greatest differences were evident for side bending, as differences in flexion were  $< 4\%$ . Comparisons to a plurimeter were also good with maximum differences measuring  $1.5^{\circ}$  at  $60^{\circ}$  bend (Perriman et al., 2010).

Such a device has shown to offer intraclass correlation coefficients of 0.96-0.88 for full range spinal movements and a series of functional tasks (Bible, Biswas, Miller, Whang, & Grauer, 2010; Perriman et al., 2010). Similar levels were reported during spinal movement whilst sitting and flexion/extension (Dolan & Green, 2006; Paquet, Malouin, Richards, Dionne, & Comeau, 1991). Furthermore the degree of repeated measurement errors was  $0.5^{\circ}$  for

repeated slouched sitting and  $0.4^{\circ}$  for repeated standing tasks (Brumagne, Lysens, & Spaepen, 1999; Dolan et al., 2006).

The validity of this method has been demonstrated by direct comparison to lateral radiographs. End range flexion and extension measurements demonstrated a correlation of  $r = 0.48$  (extension) and  $0.65$  (flexion) (Thoumie, Drape, Aymard, & Bedoisseau, 1998) and a mean difference of  $5-6^{\circ}$ . Similar results were measured for thoracic comparisons where  $r = 0.81-0.82$  and mean differences  $3.5$  for standing measures (Perriman et al., 2010). Compared to radiographs, Bible et al (2010) found the electrogoniometer to be accurate to within  $2.3^{\circ}$  for lumbar spinal movements.

In summary the electrogoniometer offers a viable option for continuous spinal motion analysis, providing information regarding the time history of movement behaviour of the spine. The technology appears to offer reliable measurements of lumbar ROM as well as demonstrating good accuracy when compared to other devices. The validity of this method appears moderate as correlations with radiographs suggest reasonable correlation coefficients and errors less than  $5-6^{\circ}$ . The limitations may lie in ability to only measure two planes as well as a physical constraint of some sensors having the two end plates attached. Despite having a small ability to telescope, this in-extensible attachment may physically restrict larger ranges of motion on some electrogoniometer designs.

#### **2.1.1.6 CA6000 Spinal Motion Analyzer**

The CA6000 is an instrument consisting of a series of six high precision potentiometers in a linkage system where changes in relative angles of the links results in a change in the resistance of the potentiometers (Troke et al., 2001). The system is usually attached to the subject over the L1 and S1 spinous processes and data can be visualised in real time. It provides dynamic measurement of three dimensions in real time therefore offering a significant advantage over many other clinical systems.

The intra-operator reliability has been demonstrated as 0.80-0.82 for flexion/extension, 0.91-0.93 for lateral bending and 0.80-0.82 for rotation range of motion testing (Troke et al., 1996). Inter-operator reliability results have demonstrated similar ICC values of 0.81-0.92 with lateral bending offering the most consistent measurements (Troke et al., 1996). Direct comparisons to radiographs are not available within the literature however implied validity has been investigated by direct comparisons to fastrak (Mannion and Troke, 1999). These comparisons provide data which questions the validity of the CA6000. Flexion values demonstrated good correlation between the two systems but the mean difference in actual values was  $8.1^{\circ}$ . Values for extension demonstrated poor correlation ( $r = 0.33$ ) with mean difference values of  $3.1^{\circ}$ . Lateral bending and rotation values showed poor correlation between the two systems ( $r = 0.44$  and  $r = 0.21$ ) and mean differences of  $7.7^{\circ}$  and  $22.7^{\circ}$  respectively (Mannion and Troke, 1999).

In summary it appears that the CA6000 can provide real time measurement of three dimensional spinal movement with good repeatability, both intra and inter-operator. However such a system appears to demonstrate questionable validity when compared to fastrak, therefore it may be that such a device offers an index of spinal motion rather than a direct representation of spinal motion. The other limitations of such a device relate to the difficulties with attachment (Troke and Moore, 1995) and the rather cumbersome nature of the device questioning its feasibility for more functional tasks.

### **2.1.2 Laboratory based systems**

#### **2.1.2.1 Opto-electrical/Video based systems**

This system utilises multiple digital cameras which visualise reflective markers. Positions of these markers are calculated relative to a reference and changes in position are used to calculate movements of the spine. Due to the necessity of line of sight a specific environment is necessary for motion capture and this system is very expensive.

The accuracy of the system is reported by opto-electronic system manufacturers as in the region of 0.1mm. This level of accuracy is dependent on set-up parameters (Windolf, Gotzen, & Morlock, 2008) and during systematic testing ranged from 0.08mm to 0.13mm, with maximum observed error measurements at 0.42mm (Windolf et al., 2008), a finding mirrored by States and Pappas, (2006). Furthermore compared to a mechanical testing jig the difference in measured angle from an opto-electronic system was reported as 0.17-1.28° (Hassan, Jenkyn, & Dunning, 2007). In regards to spinal measurement it has been shown that such a system displayed maximum errors of around +/- 2 degrees for any axis with a root mean square error of less than 1 degree (Pearcy, Gill, Whittle, & Johnson, 1987).

The validity of such a system is not known as simultaneous capture with lumbar radiographs cannot be found in the literature. This probably reflects methodological difficulties and laboratory constraints. Implied validity is however noted in the excellent agreement between actual data acquired from opto-electronic systems with data acquired from radiographs (Pearcy et al., 1987) and the excellent agreement with electromagnetic systems (Edmondston et al., 2007; Hassan et al., 2007).

The benefits of such a system lie in the ability to use multiple markers. This enables great flexibility in selecting variables of interest. If a string of sensors are used then relative motion of different regions of the spine can be assessed along with overall spinal shape. However there is a level of complexity in using the software and data processing can be time-consuming. Moreover, there are constraints in requiring a specifically defined environment and line of sight dependencies which prevent the system from being portable.

In summary, this system offers the ability to measure three dimensional spinal movements, through time providing dynamic movement information. However the system is costly,

complex and data processing methods are time consuming. This, along with the environmental constraints, prevents it from being a viable option for routine clinical use.

#### **2.1.2.2 Electromagnetic motion tracking**

Electromagnetic systems contain two parts; a source and sensor. The source emits electromagnetic energy creating a 'field' or operating zone. This acts as the origin of the coordinate system and the sensor location and orientation is taken relative to this source (van Herp et al., 2000). Sensors are attached to the skin surface and this system is portable in that it is not confined to the laboratory environment.

Manufactures of such systems describe a static measurement accuracy of 0.04mm and 0.15° root mean square for position and orientation respectively.

Reliability measurement of electromagnetic motion analysis using testing jigs has shown good results. Hassan et al., (2007) compared results from a mechanical arm with an electromagnetic tracking system, as well as those from an opto-electronic system. Their results showed that the mean difference between the arm and the electromagnetic system was -1.23°, -0.95° and 0.37° for flexion/extension, valgus/varus and inversion/eversion respectively. Compared with the optical system differences were 0.15°, -0.32° and -0.54° for flexion/extension, valgus/varus and inversion/eversion respectively. These results show excellent similarities between the two measurement systems when attached to the moving part of the 'limb'. This however does not represent skin surface attachment.

Reliability of dynamic in vivo movements have shown good reliability for treadmill walking, with intra-trial, intra-day/inter-tester, inter-day/intra-tester reliability reported at 0.94, 0.85 and 0.77 respectively (Mills, Morrison, Lloyd, & Barrett, 2007). Similar results have been reported by Burnett et al., (1998) who demonstrated a coefficient of multiple correlation of 0.89-0.94 for spinal motion measured during repeated cricket bowling actions. Cardinal spinal

motions have demonstrated an intra-class correlation coefficient of between 0.82-0.99 (Burnett et al., 1998). Pearcy and Hindle, (1989) were able to demonstrate a root mean square error of less than 0.2 degrees for spinal movement testing.

Comparing electromagnetic systems to lateral radiograph shows a difference in measured intervertebral sagittal rotation of  $0.47 \pm 0.24^\circ$  with maximum error reported as  $0.8^\circ$  (Zhao, Yang, Zhao, & An, 2005). Whilst this level of agreement is indeed impressive, importantly the electromagnetic sensors were drilled into bone and the maximum excursion tested was on average  $5.7 \pm 2.7^\circ$  of intervertebral motion. It appears that this level of accuracy may be achievable in vivo also. Steffan et al., (1997) directly compared kirschner wires in spinous processes whilst simultaneously recording from with an electromagnetic system attached to the skin. The RMS error was reported as  $0.08^\circ$  -  $0.19^\circ$  for angular motion and 0.20-0.38mm for linear motion during cardinal plane spinal movements.

A multi-sensor system has been shown to accurately predict vertebral rotation when compared to radiographs (Yang et al., 2005) and the total measurement discrepancy has been reported as less than 5 degrees for flexion using a two sensor system (Yang, Ma, Wang, & Lee, 2008).

Commonly in spinal research two sensors are used to provide information of movement behaviour at two points, usually S1 and L1. This enables the determination of spinal movement from pelvic movement (van Herp et al., 2000). Whilst this provides information regarding the position and orientation of these two points, information on spinal behaviour between these points is not known. Researchers have attempted to overcome this with the inclusion of 3 lumbar sensors (Dankaerts, O'Sullivan, Burnett, & Straker, 2006) or even a string of sensors (Gatton & Pearcy, 1999) to provide this extra detail. This however significantly increases the complexity of analysis.



The limitations of such a system are important as they also represent the key sources of error. The source emits electromagnetic energy creating an operational zone. It is within this zone that the system is designed to operate and is a cubic area around  $220\text{mm}^3$ - $720\text{mm}^3$  from the source (Bull & Amis, 1997; Milne, Chess, Johnson, & King, 1996). If measurements outside of this region are required then increases in power output are necessary (Bull et al., 1997) or consequences are noted in position and orientation accuracy (Schuler, Bey, Shearn, & Butler, 2005). This limits its usefulness for more dynamic applications constraining the subject within these boundaries. Further limitations to the systems operating accuracy is related to the presence of metal and metal interference (Milne et al., 1996). This effect appears not to be universal across all metals explaining the conflict of effect in the literature, however when evident significantly effects measurements (Ng, Burnett, Campbell, & O'Sullivan, 2009) (Burnett, Cornelius, Dankaerts, & O'Sullivan, 2004), (Milne et al., 1996).

In summary, it appears that electromagnetic motion analysis systems are small and portable, offering reliable data and a moderate to good representation of spinal motion when compared to radiographs. They are non-invasive, can measure motion in three-dimensions during any spinal movement and if a string of sensors is used provide information about relative motions of specific regions. However there are constraints on the operating zone and it may be affected by metals limiting its application in spinal motion analysis.

#### **2.1.2.3 Radiographic**

Planar radiographs have been used to study spinal movement behaviour. They generally involve the individual adopting a position (typically end of range of motion) whilst a radiograph is taken. This image is compared to an upright image where the vertebrae are traced and displacements calculated. This is considerably time consuming and represents significant

radiation exposure for the individual. Furthermore it only enables single positions to be taken and analysis is restricted to a single plane.

These problems were overcome with the development of three-dimensional or stereo-radiography (Pearcy, 1985). This involves two simultaneous radiographs positioned orthogonally with the segmental translations and rotations being calculated. This enables three-dimensional movement to be studied but can still only resolve static postures.

The ability to measure through motion was overcome by cineradiography. This essentially consisted of multiple images captured as the subject undertook a movement (Harada, Abumi, Ito, & Kaneda, 2000). Unfortunately this method provides a large radiation dose making it unsuitable for widespread use. A low radiation alternative in the form of videofluoroscopy has been suggested. This involves the use of a fluoroscope, image intensifier and a video camera enabling 2-dimensional images of the spine to be captured on to video tape (Lee, Lee, Lee, & Kwon, 2011). These are then digitised and processed. There is still some radiation exposure questioning the suitability of this method, however the main difficulty is the time-consuming post imaging processing (Panjabi, Chang, & Dvorak, 1992).

In summary radiographs uniquely offer a method of direct visualisation of spinal kinematic behaviour. However, difficulties in image processing along with the radiation exposure make it unsuitable for routine clinical use.

#### **2.1.2.4 Inertial sensors**

Robotic, aerospace and maritime engineering have used inertial sensors to provide information on orientation and motion of rigid bodies and recent years have seen the emergence of similar technologies being utilised in the field of biomechanics.

Inertial sensor as a term can be used to describe accelerometers or gyroscopes, however the term usually relates to a fusion of tri-axial accelerometers, tri-axial gyroscopes and tri-axial magnetometers used in combination to correct for individual inherent limitations.

Accelerometers usually contain a cantilevered mass element mounted on a fixed base with strain sensitive wires attached. Increased acceleration results in increased deformation of the mass element causing change in the strain in the wires. They measure linear acceleration along the sensing axes based on the equation,  $\text{force} = \text{mass} \times \text{acceleration}$ . Orientation when static may be measured by functioning as inclinometers, measuring tilting angle with respect to gravity and it is this function that is commonly used in spinal motion analysis.

The accuracy of tri-axial accelerometers has been reported at  $1.3^\circ$  for angular error with a reproducibility of  $0.2^\circ$  (Hansson, Asterland, Holmer, & Skerfving, 2001). Compared to a rotation alignment device, a RMS error of less-than  $1^\circ$  and correlation coefficient of greater than 0.99 was achieved suggesting they are highly reliable (Wong & Wong, 2008a). Biaxial accelerometers have also been shown to be highly reliable with calculated angles differing from goniometric measures by  $0.3 \pm 0.3^\circ$  and comparison to an electromagnetic system providing a correlation coefficient of 0.99 (Wong, Lee, & Yeung, 2009). They have been successfully used to monitor activity where specific activities of daily living can be identified solely from the data train (Kang et al., 2010; Mathie, Celler, Lovell, & AC, 2004; Oshima et al., 2010). Furthermore, they have been used to measure spinal movement and posture (Aloglah, Lahiji, Loparo, & Mehregany, 2010; Breen, Nisar, & O'Leighin, 2009; Lou, Raso, Hill, Durdle, & Moreau, 2002) as well as gait in low back research (Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskiold-Samsøe, 2004; Wong et al., 2008a). A limitation associated with using accelerometers for spinal motion analysis is that they measure inclination with respect to gravity based on the pendulum principle, therefore in the upright position only two planes are measurable, meaning the third (axial rotation in this case) is lost. A further limitation is the

poor quality inclination estimation for movements with large accelerations (Luinge & Veltink, 2004).

A gyroscope is an angular velocity sensor that is based on the measurement of the Coriolis force of a vibrating device whose output voltage is proportional to the rotational velocity. Angular orientation can then be obtained from the integration of the gyroscope signals (Lee, Laprade, & Fung, 2003). Such systems have been demonstrated to reliably measure spinal motion with correlation coefficients ranging from 0.97-0.99 across all anatomical plains (Lee et al., 2003).

As discussed previously it is common place to find a fusion of tri-axial accelerometers, tri-axial gyroscopes and tri-axial magnetometers in commercially available inertial sensors. In order to obtain orientations from the angular rates of rotation (i.e. gyroscopes), integration is necessary. However gyroscopes are known to suffer from integration drift when stationary (Luinge & Veltink, 2005) which can result in large errors in orientation estimation. Various methods have been proposed to overcome this limitation, including high pass filtering (Boonstra et al., 2006), low pass filtering (Mayagoitia, Nene, & Veltink, 2002), quaternion filtering (de Vries, Veeger, Baten, & van der Helm, 2009) or kalman filtering (Lee et al., 2003). The inclusion of a tri-axial magnetometer provides the addition of heading (azimuth) information and such information can aid in the correction for drift (Favre, Chardonens, & Aminian, 2007). Another method is to use an accelerometer to measure orientation when stationary (Wong & Wong, 2008c). An algorithm is developed for detection of when the sensor is (relatively) stationary and orientation relative to gravity is taken from accelerometer data and use to correct for gyroscope drift. In modern inertial sensors it is usual that combinations of these methods are used to provide drift free orientation data.

This fusion technology has a manufacture stated accuracy of  $\pm 0.5^\circ$  for static test conditions and  $\pm 2^\circ$  for dynamic cyclic test conditions. Repeated measures reliability for measuring spinal ROM has only been tested for the cervical spine, where excellent coefficient of multiple correlation (CMC) and ICC values were found (0.96-0.98; 0.87-0.92) as well as small RMSE and mean absolute errors ( $6-7^\circ$ ;  $3-7^\circ$  for full cycle movements) (Theobald, Jones, & Williams, 2011). No such reliability studies have been conducted on the lumbar spine, however they have been successfully incorporated to measure lumbar spinal motion and posture (Wong & Wong, 2008b; Wong et al., 2008c).

Validity of measures using inertial sensors during spinal movement can be implied due to the close matching demonstrated between electromagnetic systems and inertial systems (Saber-Sheikh, Bryant, Glazzard, Hamel, & Lee, 2009) and opto-electronic and inertial systems (Wong et al., 2008c). Inertial sensor systems have been shown to match an electromagnetic system to within  $0.05^\circ$  for an artificial hinge joint and  $0.28-0.69^\circ$  for random movements of a six degrees freedom wooden jig (Saber-Sheikh et al., 2009). Inertial sensors also closely matched opto-electronic systems during three-dimensional jig testing to within  $1.5^\circ$  (Wong et al., 2008c) and in the measurement of spinal posture, with maximal errors of less than  $3.1^\circ$  (Wong et al., 2008c).

As discussed previously it is commonplace within spinal biomechanics to utilise two sensors placed over the sacrum and lumbar spine. This set-up enables lumbar motion to be derived from sacral motion, however as mentioned previously the kinematics within the region between sensors is not known. This could be overcome by the addition of more sensors and is discussed in the curvature section.

In summary it appears that inertial sensors offer a portable, small, light weight spinal posture and motion analysis option. Sensors can provide reliable kinematic information regarding

dynamic motion in three dimensions and they overcome the limitations of environmental constraints. These attributes make them very attractive for clinical use however additional research is required to determine their usefulness in LBP populations and in clinical environments.

#### **2.1.2.5 Summary**

This section has reviewed some of the available motion analysis options for the measurement of lumbar spine kinematics. It is evident that no one system can provide a solution for all motion analysis scenarios and therefore the advantages and disadvantages have been discussed and are presented in table (2.1). Currently it appears possible to measure 3-dimensional kinematics without the constraints of the laboratory environment utilising either electromagnetic sensors or inertial sensors. Electromagnetic systems have inherent limitations of small operating environments and are sensitive to the presence of metal making for careful consideration during certain applications. Inertial sensors overcome these issues regarding environmental constraints and are growing in popularity within the field of biomechanics. They have been used successfully to measure spinal range of motion, however it has yet to be determined if such systems are reliable in LBP sufferers and whether significant levels of reliability can be achieved with a clinic setting. Moreover analysis to date has been limited to range of motion and as such it is not known if this technology can be used to measure higher order kinematics, such as velocity and acceleration in LBP sufferers. It may then be possible to extend the kinematic analysis to the spatial domain. The spatial domain as used in this thesis refers to the relationship between angular velocity and angular displacement where these variables are plotted in phase space to visualise the trajectory of movement. This additional information would provide useful clinical information regarding the coordination and control of movement. Therefore in light of these additional benefits and potential applications the clinical use of inertial sensors warrants further investigation.

### **2.1.3 Measurement of lumbar curvature**

The following section will review methodologies used in curvature measurement of the spine. Rather than repeat information outlined in the previous section it will extend the discussion of previously introduced methods in order to determine the application specifically for curvature measurement as well as introduce some additional technologies used specifically for curvature measurement of the spine.

#### **2.1.3.1 Photographic**

Measurement of curvature has been completed using simple lateral photographs (Seah et al., 2011; Smith, O'Sullivan, & Straker, 2008). It is commonplace to attach a series of markers to the skin over specific spinous processes and create tangents to calculate angles which represent curvature. This method has demonstrated excellent repeated measures reliability with ICC values of  $>0.99$  for standing and sitting and standard error of measurement of  $<0.5^\circ$ . However the inter-rater reliability results suggest moderate reliability with ICC values of  $>0.49$  for standing and sitting and SEM of  $>6.3^\circ$  (Perry, Smith, Straker, Coleman, & O'Sullivan, 2008). The computation method relies on the calculation of angle for a static image only. It would be possible to conduct a series of photographs to assess dynamic movement however the number of frames per second would be small resulting in lost kinematic information as well as impacting on the processing time required to digitise the pictures.

The lateral photograph method is quick and simple to administer within the clinical setting however it is only really appropriate for static images and has inherent errors with differing operators. Furthermore unless digitisation software is acquired the angle must be manually calculated from the photographs increasing the processing time considerably.

### **2.1.3.2 Flexicurve**

The flexicurve for measurement of motion has been described above along with a discussion regarding its reliability and validity. This measurement method however has some important additional advantages that merit special consideration within this section on curvature. It has the advantage of being able to measure the entire sagittal shape of the spine. All other systems would require some element of interpolation or estimation of any gap between sensors potentially introducing error. The continuous shape also enables a regional breakdown of curvature to be possible. This may be useful as different regions, even within the lumbar spine may behave differently (Burton, 1987; Burton et al., 1989; Burton, 1986; Mitchell, O'Sullivan, Burnett, Straker, & Smith, 2008). The limitations of such a system lie in the restriction that measurements are confined to static positions only. Therefore it is not possible to investigate the time relationship with spinal curvature.

Therefore in summary it appears that the flexicurve offers a unique method to observe the entire sagittal spinal shape and provide insights into regional spinal curvature, however currently this applies to static postures only.

### **2.1.3.3 Spinal mouse**

The spinal mouse is a wheeled accelerometer that is held and traced over the spinous processes by the operator, slowly to enable the accelerometer to determine inclination in the sagittal plane. The device has been used in the assessment of spinal posture across a variety of tasks and individuals (Muyor, Lopez-Minarro, & Alacid, 2011; Watanabe, Equchi, Kobara, & Ishida, 2008; Watanabe, Kobara, Ishida, & Equchi, 2010). The intra-rater reliability has been demonstrated as moderate to good, 0.61-0.93, for repeated measures across standing and end range flexion-extension (Kellis, Adamou, Tziliou, & Emmanouilidou, 2008) with similar findings presented by other authors (Mannion, Knecht, Balaban, Dvorak, & Grob, 2004). Extension consistently demonstrates the lowest reliability values (Kellis et al., 2008).



The validity of such a device has been questioned by (Ripani et al., 2008) who determined that there was no correlation between the spinal mouse and lateral radiographs. Unfortunately no actual figures of error were presented making conclusions difficult. This study was concerned with intervertebral motion however, and therefore it is not clear how comparable whole lumbar curvature measurements collected with the spinal mouse would compare to radiographs.

The limitations of the spinal mouse lie in its inability to measure dynamic motions therefore no information regarding curvature change through time can be determined. Furthermore it is limited only to the sagittal plane.

In summary the spinal mouse may offer a reliable method of curvature measurement with additional caution placed on extension measurement, however validity compared with radiographs appears yet to be adequately determined. Its major limitation is that only static measurements are possible.

#### **2.1.3.4 Inertial sensors**

As mentioned previously accelerometers can measure inclination relative to gravity based on the pendulum principle and because of this only two planes of motion are measurable at one time. The previous section has already discussed this technology; however accelerometers offer particular advantages for measuring spinal curvature. They are very cheap and do not suffer drift problems seen with gyroscopes making them usable over prolonged periods such as for postural analysis (Breen et al., 2009). Furthermore it is possible that due to their small size and low cost that a string of sensors spaced minimally apart could be used to determine spinal curvature over the length of the spine. As mentioned previously this will have the limitations of only two planes of motion (not rotation) and a degree of data interpolation would be required to yield continuous spinal shape. It would be best suited to applications

which are either psuedostatic or involve slow movements as accelerometers provide poor inclination estimates during large accelerations (Luinge et al., 2004) but would have the potential to overcome many limitations previously discussed. To the author's knowledge, no such system has been tested to date.

In order to overcome the stated limitations of accelerometers (2 measurable planes only) inertial sensors could be used. Curvature measurement is possible using inertial sensors as inertial sensors provide measurement of absolute orientation and therefore providing at least two sensors are used, the relative orientation between the two sensors represents a measure of curvature. Such a set-up has been discussed previously and has the limitations of failing to provide information between the sensors. Therefore if a similar model to that outlined above was employed, notably a string of inertial sensors spaced minimally apart, then through a process of interpolation a continuous representation of spinal shape could be obtained. This would theoretically provide fully dynamic three-dimensional curvature measurement. To date such a set-up has not been tested, however similar set-ups have been demonstrated using electromagnetic systems (see electromagnetic section). The inertial sensor version would potentially have few environmental constraints and would be easy to use within the clinic. Currently however inertial sensors are expensive and many sensors would significantly increase the cost, making them less viable for routine clinical use until such costs could be reduced. Moreover the complexity of analysis associated with many sensors would be a limitation which could potentially be overcome for the clinician by additional bespoke operating software.

In summary it may be possible to operate a string of inertial sensors spaced minimally apart, to provide a method a yielding continual spinal shape in three-dimensions. This would result in dynamic measurement of curvature behaviour through time. To date such an application has not been tested.

### **2.1.3.5 Electromagnetic motion tracking**

The use of an electromagnetic tracking system for spinal movement has been discussed earlier in this chapter, however it offers a viable option for spinal curvature measurement and warrants further discussion here. Authors have presented the use of two sensors for measuring overall spinal curvature (Stamos-Papastamos, Petty, & Williams, 2011) and as mentioned previously this may result in lost information between the sensors. To overcome this, authors have used three sensors for regional spinal curvature measurement (O'Sullivan et al., 2006) and others have even used five sensors (Gatton et al., 1999) positioned over each spinous process. This is possible due to small sensor size and a methodology which requires flexion only, as extension would result in collision of the sensors. This methodology of multiple sensors would enable the visualisation of overall lumbar curvature with potential for segmental curvature to be defined also. Moreover this method provides three dimensional curvatures to be determined dynamically through time. This method has several limitations for clinic based application. As mentioned previously electromagnetic systems are sensitive to metallic distortion which would necessitate a large clinical environment to minimise distortion and only movements completed in the operating zone are possible. Sampling frequencies are shared among sensors meaning the greater the number of sensors the poorer the sampling frequency, potentially resulting in missed kinematic information. Such a set-up requires complex data processing which may require specifically written additional software.

Interestingly because the electromagnetic system can provide both orientation and position data it is possible to utilise a sensor as a three dimensional 'pen', digitising a frame of reference (pelvis) and tracing the contour of the spine to provide a representation of spinal shape. Such a technique has been shown to provide excellent reliability with ICC values 0.98 and standard error values of  $1.5^{\circ}$  for the measurement of sagittal lumbar curvature and 0.84 and  $0.6^{\circ}$  for lateral curvature (Singh, Bailey, & Lee, 2010). This method requires a static posture to be adopted and therefore is not appropriate for dynamic curvature measurement.

#### **2.1.3.6 Opto-electronic/Video based systems**

Video based systems offer great flexibility of application and through the potential use of multiple reflective markers positioned along the spine which can be used to draw tangents and determine spinal curvature. Such a system has been used in spinal postural analysis (Wong et al., 2008c) and can provide a continuous representation of spinal shape, dynamically through time. Nonetheless this technique requires a designated space for setup and capture, is constrained by line-of-sight between the individual and the cameras. Furthermore it is costly and requires complex and time consuming data processing. These limitations make the system inappropriate for routine clinical use.

#### **2.1.3.7 Fibre-optic sensors**

Fibre-optic technology is commonly used in medical and engineering settings to provide information on shape and visualisation of awkward environments and more recently fibre-optic sensors have been used in animation, particularly for the video game industry. Body suits housing multiple sensors are worn and used to reconstruct a three-dimensional image of the wearer. As this is often the aim in biomechanics research attention has turned to whether similar technology can be applied to this field.

The fibre-optic sensor systems pass a light from one end to the other. Light flow through the sensor is modulated by the degree of bend and therefore light intensity can be used to calculate degree of bend and subsequent orientation. If a string of sensors registering this change in light flow were spaced minimally apart then through a process of smoothing, the output can produce a curve along the entire ribbon of sensors.

A commercially available fibre-optic system (Shapetape<sup>TM</sup>, Measurand, New Brunswick, Canada) has been used for creating and manipulating curves within computer science (Grossman, Balakrishnan, & Singh, 2003). Moreover it has been compared with an opto-trak

motion analysis system using an artificial joint. Unfortunately due to experimental setup the system was unable to detect medial and lateral movement. This is due to the fibre-optic ribbon being flat and therefore unable to bend in the lateral direction unless first twisted. Results for twisting showed that the fibre-optic ribbon significantly underestimated the amount of twist due potentially to the stiffness within the sprung steel core providing resistance to twist, or due to a small degree of movement at the attachment point of the sensors. Bending through its long axis was more promising with results displaying consistent overestimation of angles as reported by opto-trak, with a magnitude of difference between  $5.1 \pm 2.4^\circ$  to  $6.9 \pm 1.2^\circ$  for angles set at -10 to 50 degrees (Morin & Reid, 2002).

In vivo use has shown promising results also. A specially adapted glove has been used to compare hand manipulation during a simulated surgical task, with the aim of comparing expert with novice. Such a system was clearly able to discriminate between operators as well as being able to identify specific tasks just from the data train. (King, Lo, & Yang, 2005). Furthermore Bell and Stigant, (2007) demonstrated the use of two fibre-optic goniometers to measure spinal and hip motion. These goniometers unfortunately provide movement data for one movement plane only, but can successfully describe dynamic motion about the plane of interest. It is worthy of note that again movement and task identification was possible based on the data train, removing the need for additional video camera analysis. Significant errors were associated with re-attachment of such a system, in the region of  $11.6^\circ$  and  $14.2^\circ$  for lumbar spine and hip respectively during standing and full flexion.

In summary, fibre-optic technology is relatively new in the field of biomechanics but may be a viable option for curvature and motion measurement. The technology appears to be able to detect movement and provide data trains representative of various functional tasks. More specifically, the fibre-optic ribbon (Shapetape<sup>TM</sup>) is able to provide a continuous representation of shape. This could therefore potentially describe the curvature of the spine for any region of

interest throughout movement. Currently the system struggles with lateral bending motions due to the product design and its reliability and validity to date has not been tested. Despite this, the system potentially has the ability to measure curvature, dynamically at fast sampling frequencies providing information about sagittal spinal shape. It can do this without reliance on a laboratory environment and at relatively low cost. Furthermore, due to the string-of-sensors design it may be possible to register regional changes in spinal movement behaviour. In light of these potential benefits the fibre-optic ribbon may warrant further investigation as a clinic based option for curvature measurement.

#### **2.1.3.8 Summary**

It has become common place for clinicians to measure lumbar curvature in the assessment of the lumbar spine (Dankaerts et al., 2006; Dankaerts et al., 2009; Scannell & McGill, 2003; Youdas, Garrett, Egan, & Therneau, 2000; Youdas, Hollman, & Krause, 2006). Lumbar curvature has been linked to LBP (Dankaerts et al., 2006) and has been shown to influence the activation levels of the trunk muscles (O'Sullivan et al., 2006; O'Sullivan et al., 2002) as well as affect the line-of-action of the paraspinal muscles (McGill, Hughson, & Parks, 2000). Furthermore curvature determines the degree of load sharing across the motion segment (Adams & Hutton, 1980a; Adams et al., 2000; Wilke, Neef, Caimi, Hooglan, & Claes, 1999) and affects the load tolerance of particular anatomical structures, altering the vulnerability of the back to injury (Gallagher et al., 2005). Therefore the measurement of lumbar curvature is important in the understanding of lumbar function.

It is clear from this review that there are many options available; however it is evident that methods of measurement share common limitations, such as the inability to measure dynamic lumbar curvature (see table 2.1). Photographic, flexicurve and the spinal mouse all offer the ability to provide a representation of the entire spinal contour providing shape information

about any region of interest, however this is limited to static postures only. These systems therefore cannot inform about the time history of changes in curvature. More complex laboratory systems, such as electromagnetic sensors or inertial sensors, are able to measure curvature through time however they have environmental or cost constraints. It is commonplace to use just two sensors, one over L1 and one over S1 thereby modelling the lumbar spine as one 3-degrees of freedom joint (Burnett et al., 1998; Lee et al., 2003; Stamos-Papastamos et al., 2011). This methodology is unable to acquire curvature information between the sensors providing an incomplete picture of lumbar curvature. Dynamic measurement of curvature would enable the sequencing of curvature change to be investigated. Sequencing of lumbar curvature has not been studied in depth in relation to LBP, which may reflect the degree of technical difficulty of measuring regional or segmental spinal curvature. Furthermore, if a continuous representation of curvature was obtained then theoretically sequencing of curvature change could be studied for any region of the spine. There is therefore a need to develop a system that is capable of resolving continuous spinal curvature dynamically through time. The attributes of the fibre-optic ribbon suggest that this may be possible at a cost which is reasonable to clinicians. It could in theory be able to resolve spinal curvature during motion for any point of interest along the ribbon's length. The result would be a quick and easy way of measuring curvature for any region or even segment along the spine in one data capture. Furthermore the system is fully portable and is not affected by environmental constraints making it ideal for use within clinical environments. Therefore in light of these additional benefits and potential applications the clinical use of the fibre-optic ribbon warrants further investigation.

Table 2.1. Summary of curvature and motion analysis review.

	Non-invasive	Portable	Reliability IntraCC	Reliability Inter CC	Validity	Dimensions	Curvature	Comments
Skin distraction	✓	✓	NK	NK	0.97*	1	x	Static only
Inclinometer	✓	✓	F: 0.79-0.91 E: 0.60-0.75 SF: 0.83-0.92	0.74-0.88 0.42-0.55	0.73-0.80* 0.75-0.15*	1	✓	Static only
Finger-tip-to-floor	✓	✓	F: 0.95-0.98 E: 0.78 SF: 0.91	0.98 0.79 0.98	NK	1	x	Static only Construct validity floored
Flexicurve	✓	✓	F: 0.95 E: 0.97	0.82-0.99	<1° *	1	✓	Static only
Electro-goniometer	✓	✓	0.88-0.96	NK	F: 0.65 E: 0.48 <6-2.3°	2	✓	Only two sensors
CA6000	✓	✓	0.80-0.93	0.81-0.92	EM 3.1-22.7	3	✓	Cumbersome
Photograph	✓	✓	S: 0.99	S: 0.49	NK	1	✓	Static only
Spinal mouse	✓	✓	F: 0.93 E: 0.61	NK	NK	1	✓	Static only
Video	✓	x	± 2°	NK	EM ± 2° jig	3	✓	Dynamic Environmental constraints Expensive
Electromagnetic	✓	✓#	0.89-0.94	NK	<5° total lumbar	3	✓	Dynamic Environmental constraints
Inertial	✓	✓	0.97-0.99 (gyro) 0.96-0.98 (Cx)	NK	EM<0.7°	3	✓	Dynamic
Fibre-optic	✓	✓	NK	NK	V 5-7°	1 <sup>§</sup>	✓	Dynamic

\* Compared to radiograph; F, Flexion; E, Extension; SF, side-flexion; S, standing; EM, compared to electromagnetic; #, environmental constraints; gyro, gyroscopes used for measurement; Cx, cervical spine; V, Video-based opto-electronic system; <sup>§</sup> Capable of 3 dimensional measurement but poor functionality in other planes; NK, not known



## **2.2 IS PAIN THE CAUSE OF ALTERED LUMBAR KINEMATICS AND ELECTROMYOGRAPHY IN BACK PAIN SUFFERERS?**

### **2.2.1 Introduction**

Low back pain (LBP) is a major health and socioeconomic burden, and a leading cause of disability (Frymoyer, 1988). LBP sufferers often display changes in the biomechanical behaviour of the trunk including, but not limited to, alterations in movement patterns and muscle activity. The nature of these alterations remains poorly understood. Lumbar kinematics and muscle functions have long been a key focus within LBP and are vitally important as they are strongly associated with risk of LBP onset and LBP reporting (Norman et al., 1998; Stevenson, Weber, Smith, Dumas, & Albert, 2001; Van Nieuwenhuyse et al., 2004) and are included in governmental guidelines on safe handling and impairment measurement (Cocchiarella et al., 2000; Waters, Putz-Anderson, Garg, & Fine, 1993).

Kinematic alterations have commonly been identified in sufferers of LBP (Marras, Davis, Ferguson, Lucas, & Gupta, 2001; Marras et al., 1999; Marras & Wongsam, 1986; Shum, Crosbie, & Lee, 2005a; 2005b; Wong et al., 2004) and the higher order kinematics, such as velocity and acceleration, strongly correlate with the loss of functions and disability (Marras et al., 2001; Marras et al., 1999; Marras et al., 1995; Marras et al., 1986; Novy, Simmonds, Olson, Lee, & Jones, 1999; Shum et al., 2005a; 2005b; Shum, Crosbie, & Lee, 2007a). There have also been attempts to study changes in muscle behaviour (Hodges & Moseley, 2003b; van Dieen, Selen, & Cholewicki, 2003) however no universal consensus exists. Attempts to explain changes in amplitude of muscle activation have been complicated by two conflicting models, the pain-spasm-pain model (Travell, Rinzler, & Herman, 1942) and the pain-adaption model (Lund, Donga, Widmer, & Stohler, 1991). The pain-spasm-pain model predicts that pain will induce muscular hyperactivity or spasm which would in-turn cause

pain. However, the pain-adaption model predicts that when pain is present, muscle activation patterns are altered according to their particular function.

Previous reviews (Marras et al., 1999; van Dieen et al., 2003) have focussed on reporting the alterations or impairments observed in LBP sufferers and more recent reviews have analysed effect sizes, beginning to outline which alterations are associated with LBP (Geisser et al., 2005). Reviews however have not been able to address the causative mechanisms, therefore a critical review of the current understanding of the issue of pain and lumbar kinematics and muscle function is required.

Management strategies often involve targeting pain relief to alter spinal biomechanics and functions (Jette et al., 1994). If the aim of treatment is to restore the biomechanical behaviour of the spine then it is imperative to identify the underlying cause or mechanism responsible for the biomechanical changes so that clinical management can be rationalised. The question remains as to whether pain drives movement and muscle changes. Therefore, this review aims to explore the concept of pain driven changes in lumbar kinematics and muscle functions examining the experimental pain models employed in this area of research. It will discuss how this information can be used in the development and justification of clinical management models aimed at restoring the biomechanical behaviour of the spine. However, it should be acknowledged that many other variables have been suggested to cause changes in biomechanical functions, including, but not limited to, spinal stiffness (Lee, Tsung, Tong, & Evans, 2005), and fear of movement (Thomas & France, 2007; Thomas, France, Sha, & Vander Wiele, 2008), however by far the most commonly cited is pain (Hodges et al., 2003b). It is pain that will form the basis of this review, concentrating on the immediate effects of pain induction and pain relief on biomechanical functions.

### **2.2.2 Methods**

To be included in this review studies needed to meet the following criteria. Articles needed to investigate either the effects of experimentally induced pain or that of experimental pain relief related specifically to the low back region. The review was limited to these methodologies as it is thought they studied the effects of pain as a separate variable. All measurements had to be completed immediately and include either lumbar kinematics or muscle function. Immediate measures only were selected in an attempt to maximize the impact of altering just one variable; pain. Searches were completed of Medline 1948-2009 (English language only) using a variety of terms including LBP, experimentally induced pain, pain-relief, kinematics and biomechanics, along with reference lists of retrieved articles. Fifteen studies matching the above criteria were retrieved and are presented in tables 2.2 and 2.3. A systematic review of the methodology of these studies was completed using a modified version of the criteria list as suggested by Downs and Black (1998) (see appendix 1.0) and the results are presented in the appropriate column of tables 2.2 and 2.3. The criteria list was modified by removing item 27 as this item dealt with the follow up of patients which was deemed not applicable to these experimental studies. Effect size calculations were also carried out, the results of which are presented in table 2.2 and 2.3. It has been suggested that an effect size of 0.2 is small, 0.5 is moderate and  $>0.8$  is large (Cohen, 1988; 1992).

### **2.2.3 Results**

Studies utilising the method of experimentally induced pain share common methodologies and are at risk of common threats to validity. All studies failed to report potential confounding issues (question 5) and adjust for any of these issues (question 25). All studies failed to clearly report any adverse effects or an absence of adverse effects. Baseline characteristics are not often reported as initially the subjects are 'normal'. These studies were all carried out in

laboratory environment therefore all studies score poorly on question 13. Due to the nature of the experimental method blinding the subject and randomising is not always possible, seen in the scoring of questions 14, 23, 24 and 25. One potential source of bias is the failure to blind data processing for trial type (question 15).

Table 2.2 outlines the results of the systematic methodological analysis for studies of experimental pain relief with most studies struggling to control confounding variables. Poor reporting of adverse effects is visible by low scores to question 8. This line of research enquiry often relies on a convenience or consecutive sample providing a threat to external validity. Results of effect size calculations are presented in tables 2.2 and 2.3. Thirteen effect size calculations were possible for experimentally induced pain studies, where the largest effect size was obtained for changes in onset time of deep lumbar multifidus during shoulder flexion. Large effect sizes were also evident for onset of transversus abdominus during shoulder flexion, thickness of transversus abdominus during abdominal hollowing and mean amplitude of emg for erector spinae during gait.

Three effect sizes were calculated for studies employing experimental pain relief, with small to moderate effect size for Sorensen test improvement displaying the largest effect size.

## **2.2.4 Discussion**

### **2.2.4.1 Methodological Analysis**

The reporting of confounding issues is important to understanding the factors influencing the results, however despite the lack of reporting in experimentally induced pain models, it could be argued that the method of using 'normals' is a good way of controlling or minimizing the impact of these confounding variables. It is however not clear whether pre-existing traits impact on the experience of induced pain and therefore affect the results. The importance of confounding variables in experimental pain relief studies should not be understated. It is well known that LBP populations are far from homogenous making it very difficult to control for

these confounding variables, questioning the true meaning of the results. Due to the multifactorial nature of LBP, studies of this nature using a sample of convenience or consecutive samples are likely to contain mixed sub-groups. The reporting of adverse effects is important to determine the safety profile of specific interventions, which is imperative in pain relief trials if the interventions are to be advocated. Poor reporting of these factors in both groups means the safety of the experimental method of inducing pain is not clear and the clinical usefulness of pain relief strategies employed is not clear. The lack of blinding of investigators is commonly observed in these studies; however the importance of such is questionable as in this type of quantitative research processing methods are often automated by computer programs which remain consistent throughout the analysis.

#### **2.2.4.2 Experimental Pain Models**

As can be seen from table 2.2, three studies utilising induced pain investigate the kinematics of gait. Clear attenuations in gait have been demonstrated in LBP sufferers (Keefe & Hill, 1985), however induced pain failed to alter relative phases of gait (Arendt-Nielsen, Graven-Nielsen, Sværre, & Svensson, 1995) or trunk coordination in the transverse plane (Lamoth et al., 2004). It was deduced that the key determinant of relative phase trunk coordination was in fact walking velocity not pain.

In contrast, Moe-Nilssen and colleagues (1999) were able to show that experimentally induced pain did indeed attenuate walking kinematics, but these attenuations were not in the relative phase couples or phases of gait but were actually in walking velocity itself. As can be seen in table 2.2, previous studies had controlled for walking velocity therefore masking this temporal kinematic change. Moe-Nilssen et al., (1999) asked the subject to walk at a velocity of their choosing which varied on prompting. This protocol enabled clear reductions in overall velocity to become visible, something strongly associated with LBP sufferers (Keefe et al., 1985; Lamoth, Stins, Pont, Kerckhoff, & Beek, 2008; Lamoth, Daffertshofer, Meijer, & Beek, 2006; Lee, Simmonds, Etnyre, & Morris, 2007). This change results in more 'in-phase' relative

coupling of the trunk, (Lamoth et al., 2004) less spinal motion, lower joint forces and a situation closer to static loading for the lumbar spine (Callaghan, Patla, & McGill, 1999). It is clear that if equivocal speeds are investigated, induced pain does not alter lumbar kinematics during gait, however when verbal cueing for speed i.e. preferred speed or the interpretation of 'as fast as possible', alterations are seen in that slower speeds are adopted regardless of the speed requested. Interestingly this is not only seen during gait, it is also evident during forward bending. Zedka and colleagues (1999) showed that induced pain altered lumbar kinematics during forward bending by reducing the velocity of movement, along with 10-40% reduction in range of motion. Unfortunately effect size calculations were not possible due to poor reporting of actual numbers. However subjects managed to move at equivocal velocity and range when prompted by an accelerometer further suggesting that pain in some way effects the selection of movement velocity. It appears that induced pain sufferers are able to achieve equivocal speeds but 'choose' not to. This suggests a resetting of the 'velocity' control from the nervous system when the body is 'in pain' as it is not task specific (Simmonds, 2006), a finding also evident in clinical LBP (Marras, Lewis, Ferguson, & Parnianpour, 2000b; Marras et al., 1986; Novy et al., 1999). It seems logical that this may be a strategy to reduce loads on sensitive tissues, as greater velocities are known to result in greater spinal loads (Callaghan et al., 1999; Cheng, Chen, Chen, & Lee, 1998). Interestingly however it has been noted that when asked to move as fast as possible, clinical LBP sufferers are often unable to achieve equivocal speeds to matched controls (Lee, Simmonds, Novy, & Jones, 2000; Marras et al., 1986). The true reason for this remains unclear; however it may represent an unwillingness to evoke pain (if pain driven) or a loss of functional capacity of the lumbar spine suggesting a mechanism other than pain may be important.

In order to study the effects of experimentally induced pain on trunk muscle function electric shock and hypertonic saline have been utilised with a focus on paraspinal and abdominal muscles. Voluntary arm movements coupled with a painful stimulus (electric shock) show a gradual process of activation change in both the lower abdominals (transversus

abdominus/internal oblique) and, although less dramatic, in external oblique (Moseley & Hodges, 2005). This gradual process of reduced activation of the lower abdominals has been argued to represent an adaptation towards an alternate trunk muscle strategy and appears to suggest that pain (experimentally induced) may have the capacity to drive change in trunk muscle activation strategies. Moreover, following the pain-movement coupling, a period of uncoupling was completed where a return to the original activation patterns were observed, further suggesting pain may be the key instigator for these changes. Unfortunately no real numbers were reported making the interpretation of the magnitude of effect or an effect size calculation not possible. Importantly this method of inducing pain enables the observation of non-immediate changes whilst minimising the impact of other potential variables. Saline injection is very short lasting and is unable to study the subtle changes occurring over time, whereas electric shock can be delivered over a longer time period allowing the lumbar system time to adapt to the noxious stimulus. Adopting similar methodology, but using saline injections to induce pain, EMG results have shown a consistent pattern of reduced or delayed activation during voluntary arm movements (Hodges, Moseley, Gabrielsson, & Gandevia, 2003a; Moseley, Nicholas, & Hodges, 2004). These findings are believed to mirror that of small clinical LBP trials (Hodges & Richardson, 1996; , 1999). These studies follow a similar protocol involving a static posture onto which the subject performs a rapid shoulder movement. This relatively simple task relies on the adoption of identical postures throughout due to the effect of small postural changes on trunk muscle EMG (Claus, Hides, Moseley, & Hodges, 2009; O'Sullivan et al., 2006; O'Sullivan et al., 2002), however no postural measures were conducted to ensure this criteria was controlled. It is important to note that the impressive effect sizes reported for these studies represent a measure of statistical assurance rather than magnitude of effect. The magnitude of difference for muscle onset relative to deltoid, compared with controls was 28ms and 10.3ms for transversus abdominus (Hodges et al., 2003a; Moseley et al., 2004). Interestingly isotonic saline (not painful) also had a significant impact on the latency of onset of transversus abdominus compared with controls, with the magnitude of delay in

onset being 5.2ms, along with a delay of 25.1ms for superficial multifidus suggesting factors other than pain may at least be of some significance (Hodges et al., 2003a). Furthermore, when studying truly comparable experimental conditions, namely isotonic saline with hypertonic saline, the onset difference was 5.1ms for transversus abdominus. These results are further complicated by the use of visual inspection to detect EMG onset rather than an automated computer algorithm resulting in a potential source of bias (Allison, 2003; DiFabio, 1987; Hodges & Bui, 1996). The clinical significance of such a small delay in muscle onset is not well understood.

It has been previously reported that trunk kinematics are affected by the induction of pain and this often takes the form of reduced velocity of motion. In the analysis of the experimental method used by the above studies, reported kinematic data regarding the moving arm is often insufficient. Moseley and Hodges, (2005) report only deltoid EMG parameters, whereas Moseley et al., (2004) and Hodges et al., (2003a) only report p-values for peak acceleration. The reporting of shoulder movement velocity is critical as this has a large effect on trunk muscle onsets during this experimental protocol (Hodges & Richardson, 1997). As the magnitude of such an effect has been reported as 294ms delay in transversus abdominus onset for slow limb movement compared with 19ms for preferred speed, it is unclear if the delays outlined by these studies are the result of pain or are the manifestation of minor alterations in shoulder movement velocity.

Similar findings have been observed using ultrasound imaging where changes suggestive of reduced activation were observed for transversus abdominus, during abdominal hollowing and lumbar multifidus during prone limb raising (Kiesel, Uhl, Underwood, & Nitz, 2008). The findings were further replicated using a novel functional magnetic resonance imaging method, displaying changes suggestive of reduced activation of both lumbar multifidus and erector spinae during a Sorensen manoeuvre (Dickx et al., 2008). These studies tested activity in a



static condition removing any velocity deviations which may confuse interpretation, suggesting that pain may indeed attenuate changes in muscle activation. However, it should be remembered that thickness is a morphological parameter which may not directly reflect muscle function and that there are inherent difficulties with accurate re-positioning of the ultrasound probe leading to significant errors in thickness measurement, around 6-10% for multifidus cross-sectional area (Stokes, Rankin, & Newham, 2005).

Unfortunately these studies only examined very simple activities such as arm movement and fail to provide answers as to the effect of experimental pain on muscle activation during more functional tasks. Studies on functional tasks have provided conflicting results due to methodological differences in analysis techniques and tasks completed (see table 2.2). During lumbar flexion, following right sided erector spinae muscle injection, a loss of bilateral flexion relaxation response in the erector spinae was observed, something highly correlated to LBP sufferers (Geisser et al., 2005; Watson, Booker, Main, & Chen, 1997) along with reduced activation during the return from flexion, a time normally associated with high activation levels (Zedka et al., 1999). However, when the subjects were guided to complete the flexion motion identical to the painless trial (equivocal range and velocity), only the injected side displayed alteration (Zedka et al., 1999). Therefore it appears that the spine still has the functional reserve to achieve more selective muscle activation patterns in these temporary pain states, but an alternative strategy is adopted. This could reflect an attempt to avoid asymmetrical loading associated with unilateral muscle activity or a more gross reaction where the nervous system switches to function in an altered 'pain mode' regardless of the location of pain. This finding has also been seen during gait (Arendt-Nielsen et al., 1995; Lamothe et al., 2004). This suggests increased activation at a time normally associated with little or no activity, along with decreased activity during a time normally associated with large activation. These superficial muscle activation changes appear to mirror that of the pain-adaptation model (Lund et al., 1991); however these biomechanical changes are only evident during self-

selection of velocity during functional tasks. The reason for these changes is unclear. The choosing of slower functional movements may be the cause or effect of muscle activity changes. These changes may result in a reduction of movement velocity or a reduction in movement velocity may cause an increase in superficial muscle activity. Furthermore these changes may be the cause or result of changes in the deep trunk musculature. Reductions in movement velocity and alterations in superficial trunk muscle activity may result in an alteration in the deep muscle activation requirements for highly specific and coordinated activities.

Caution should be exercised when extrapolating these results to clinical pain as experimentally induced pain fails to closely mimic clinical pain. The resultant pain is always constant in nature with very little deviation except a gradual reduction over time. The pain source in these subjects is likely to be the nociceptors within the muscle, irritated chemically and locally, the presence of which in true clinical LBP is not known. It is also noteworthy that these experiments often involve injections at the level L3 (see table 2.2) whereas clinically the highest incidence in LBP is known to be the two lower levels. Furthermore, due to the transient nature any alteration in central pain processing will be minimal, as will the levels of concern regarding uncertainty about the personal meaning of their LBP.

In summary it seems from the results of these studies that experimentally induced pain results in an automatic attenuation of range of motion and reduction in movement velocity. Induced pain results in an elevation of superficial muscle activity which is often bilateral, if autonomous selection of velocity and range are permitted. This may represent a protective response and act as a method of reducing range and velocity of motion. Functional tasks appear to display EMG changes which correspond to an increase in activation at a time normally associated with little or no activity in the ipsilateral erector spinae. This is consistent with aspects of the pain-adaptation model, which unfortunately is not seen consistently in LBP sufferers (van Dieen et

al., 2003). However not consistent with this model is the small delay in onset activation seen in the deep muscles during shoulder movements. It is true some of these changes appear to mirror those present in LBP sufferers clinically and may suggest therefore that they represent pain induced changes. It is clear that the musculoskeletal system is capable of achieving equivocal ranges and velocities if guided, along with more selective EMG patterns, suggesting an alteration in the nervous system control of movement parameters when in an experimentally induced pain state. Future studies should take care to identify and control for confounding variables and adjust the method or analysis accordingly.

#### **2.2.4.3 Pain relief models**

Obviously in order to overcome the limitations of the experimental pain model a painful clinical sample could be studied and the effects pain relief investigated. The effect of pain relief on muscle testing has shown mixed results (see table 2.3). Using the Biering-Sorensen test with a group of chronic LBP sufferers, Rashid and colleagues (2003) showed that intravenous opioid increased performance some 28% compared with placebo. Furthermore Jarzem et al., (2005) displayed 15% gains in maximal isometric lifting capacity following TENS induced pain relief, compared with sham TENS. Conversely, Holm et al., (2000) utilising bilateral zygapophyseal joint injections to induce pain relief in chronic sufferers, failed to detect a significant change in muscle function as measured by an isokinetic through-range resistance task. It is important to note however that the researchers struggled to achieve a significant reduction in pain in all three studies with visual or verbal analogue scale (VAS) changes ranging from 0.9 to 21.2mm. This may reflect the underlying pathological changes or subtle differences in baseline characteristics (see table 2.3), or the inherent difficulty of reducing pain when using maximal muscle test outcomes. This notion is further complicated by psychometric testing, suggesting the Biering -Sorenson test examines pain tolerance and motivation, rather than muscular endurance (Novy, Simmonds, & Lee, 2002) and that fluid

injected into joints may have an inhibitory effect on muscle activation (Spencer, Hayes, & Alexander, 1984).

Using zygapophyseal joint injections in chronic LBP, Lilius et al., (1989) displayed immediate improvements in flexion and rotation range of motion following an 18.3mm reduction in VAS (see table 2.3). This ROM change was not universal as no difference was seen in extension or lateral flexion. Similar results were obtained when using TENS for pain relief which resulted in small gains in flexion and extension ROM. However despite significant results, actual change in ROM were either not reported (Jarzem et al., 2005) or very small, 1.4 cm and 1.5 degrees for flexion and rotation respectively (Lilius et al., 1989). Despite the fact that no attempts were made to correlate pain reduction and functional change, the findings do suggest that minor pain relief may be capable of changing lumbar kinematics; however the clinical significance of the magnitude of change could be questioned.

It seems evident that the voluntary selection of speed is not only affected by experimental and clinical pain, (Simmonds & Rebelo, 2003; Zedka et al., 1999) it may be affected by pain-relief. In studies investigating velocity it is evident that pain relief is capable of increasing movement velocity (Davis & Kotowski, 2005; Simmonds et al., 2003). Unfortunately, Davis and Kotowski, (2005) failed to present actual pain data or describe the interventions (massage, chiropractic, physical therapy or acupuncture) making the correlation between pain relief and kinematic change difficult. In their study investigating time taken to complete a repeated sit-to-stand task at three different self-selected speeds, Simmonds and Rebelo, (2003) were able to show that the fastest speed achieved by the chronic LBP group was equivocal to the preferred speed of the control group. These results suggest that chronic LBP subjects may actually be unable to achieve the same movement speeds, something observed previously in a clinical population (Marras et al., 2000b; Marras et al., 1986; Novy et al., 1999; Simmonds et al., 2003). Unfortunately it is not clear whether the task evoked pain at the time of testing, resulting in an unwillingness to move faster due to pain provocation or whether the LBP sufferer just doesn't have the functional capacity to produce the same speed, due to some unknown mechanism.

However, following pain relief induced by a superficial heat wrap, the LBP group significantly increased their sit-to-stand speed, interestingly only at the preferred speed. Therefore pain relief may have resulted in an adaptation of the lumbar spine system through its neural control causing a shift away from its 'pain' setting (Simmonds, 2006) at the preferred speed. This may to some extent explain the positive clinical effects associated with the application of topical heat (Nadler et al., 2003a; Nadler et al., 2002; Nadler et al., 2003b). Moreover, due to the specific nature of the changes it could be argued that the effects were unlikely due to changes in the deeper tissues known to be influenced by heat (Bass et al., 2007), but rather due to the simple relief of pain.

In summary, there is some evidence to suggest that pain relief results in an automatic increase in movement velocity when the self selection of speed is permitted during functional tasks. It appears feasible that pain relief may be able to alter the ROM but the true magnitude of effect suggests questionable clinical significance. It remains unclear as to how pain relief affects muscle function as the results are variable. Isometrically, performance is improved; however through range strength testing shows no effect. It is questionable how an improvement in maximal muscle testing relates to functional daily tasks, which seldom require the full capacity of the lumbar muscles and no EMG studies have been conducted. Future research should concentrate on careful selection of inclusion criteria in an attempt to create a relatively homogenous sample and minimise confounding variables, along with careful reporting of adverse effects to determine the clinical application of the specific pain relief strategies investigated.

### **2.2.5 Conclusion**

This section has provided a contemporary review of the most current understanding of the relationship between pain and biomechanical functions of the trunk. It has identified several

important biomechanical features associated with LBP. These relate to the altered kinematic patterns and associated changes in muscle functions during self-selected tasks. The reason for these changes remains unclear. There is clearly a relationship between movement velocity and activity changes in the superficial or deep trunk muscles and we do not know their cause and effect or how the deep and superficial muscles interact. It is suggested that the CNS may require more 'time' to control and coordinate movements, modify muscle activities and thus minimise pain provocation. On the other hand, it is entirely possible that these adaptations could be detrimental to spinal health and function. Pain may induce the cascade of movement and muscle changes which represent sub-optimal function, providing an ongoing mechanism for symptom provocation.

Understanding the mechanisms behind these alterations would significantly enhance the understanding of their functions. Clinical management may involve interventions which optimise pain relief or target some other mechanisms causing biomechanical change.

Therefore further studies are required to isolate and identify individual mechanisms and to test their influence on trunk muscle functions and kinematics. Only once the key mechanisms underpinning the alteration in function in clinical LBP populations are identified can clinicians and researchers alike begin to employ rational and specific interventions resulting in the restoration of normal biomechanical behaviour of the trunk.

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**Summary**

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**What is known?**

Acute and chronic LBP sufferers display altered biomechanical functions, including movement and muscle behaviours.

Experimental studies replicate some of the changes evident in LBP sufferers.

**Limitation of what is known?**

Underlying mechanism causing these changes is not known.

Experimental models suggest pain may be one key variable of interest; however this is not clear due to methodological limitations.

**Further work.**

Future work is needed to define the roles of various mechanisms in instigating the observed changes in biomechanical functions.

Clinical management strategies should then be developed and verified to address the underlying mechanisms in clinical LBP populations.

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Table 2.2. Experimentally induced pain studies

Authors	Subjects and Task	Induced pain	Outcome measures	Results	Comments	Effect size	Downs and Black missing criteria
Arendt-Nielsen et al.,(1995)	10 male Age: 23-30 mean: 25 No known LBP Treadmill walk 4.0km/h for 1 minute.	0.5ml hypertonic (5%) saline. Depth: 1.5cm Location: 4.0-4.5cm lateral to L3 on the right	Gait phases. Mean amplitude of non-amplitude normalised EMG profiles (mean EMG). Ratio between mean EMG right and left (ratio EMG). Mean EMG of peak activity at double stance phase (peak <sub>50</sub> ). Mean EMG in contra-ipsilateral swing phases. Surface EMG placement: Lateral to Th 12, L2, L2 lat, L4 bilaterally.	Gait phases not affect by pain. Mean EMG increased 8.5%. No change in ratio EMG. Peak <sub>50</sub> decreased 7.3%. Mean EMG during contra- and ipsilateral swing increased 15.1% and 19.2% respectively. No correlation between EMG and VAS (R=0.53)	Mean VAS 5.4±2.3. EMG ipsilateral to pain at L2 showed most significant changes.	1.2* (mean EMG) 0.81* (peak <sub>50</sub> ) 1.04* (contra-lateral swing) 1.03* (ipsilateral swing)	3,5,8,11,12,14,15, 21-25
Zedka et al.,(1999)	4 males and 1 female Age: 20-55  Flexion (constrained lower limbs).	5% Saline infusion  Depth: 4cm Location: 3cm lateral to L3 on the right.	Trunk displacement. Velocity of motion. EMG amplitude.  Surface EMG placement: Lateral to L3 bilaterally Stretch reflex of erector spinae muscles.	Pain decreased trunk displacement (60-90%) and reduced velocity. EMG amplitude during flexion unchanged. Loss of FRR bilaterally. EMG amplitude decreased during return from flexion bilaterally.	Mean VAS 5.3±0.8 During painful movement guided for velocity and displacement, EMG on contralateral side identical to painless trial and EMG on ipsilateral side displayed loss of FRR and decreased activation on return	NA	3,5,8,10-12,14,15,21-25
Lamoth et al.,(2004)	8 male and 4 female Age: 18-25	0.5ml of 5% saline Depth: 30mm	Relative phase coupling. EMG amplitude and patterns	Pain had no effect on phase coupling. Elevated right EMG during	Mean VAS 6.1±1.9. No correlations between pain	NA	5,8,12,15,22,24,25



	mean: 21 No Known LBP. Treadmill walking 1 minute at 2.2, 3.8, 4.6, 5.4km/h.	Location: 65mm lateral to L3 on the right.	Surface EMG placement: 3cm lateral to Th12, L2, L4 bilaterally	ipsilateral swing for all locations and for L2 during contralateral swing phase. Elevated left EMG amplitude for ipsilateral swing phase.	intensity and EMG findings.		
Moe- Nilssen et al.,(1999)	3 male and 19 female Age: 20-49 mean: 26 Walking at five self adjusted speeds.	1ml 6% saline Depth: 15.2mm Location: 34mm lateral to Th12 or L1 on the left.	Trunk acceleration during walking: AP axes, ML axes, Vertical axes	Significant attenuation of acceleration in AP and ML axes when pain evident. Not for vertical axes.	20 only reached significant pain levels. Mean VAS 6.1±1.49. 15 of 20 showed correlation $R^2$ 0.36- 0.89 between pain and gait changes.	0.75* (AP accel) 0.27* (vertical accel) 0.75*(medio-lateral accel)	5,8,9,12,14,15,22- 25
Moseley et al.,(2004)	5 male and 3 female Age: 32±7 No known LBP. Single shoulder flexion (standing)	1-1.5ml 5% Saline Depth: 30mm  Location: 50mm lateral to L4	Temporal and Spatial parameters of EMG (related to reaction times).  Intramuscular EMG: deep LM, superficial LM, OE and OI.	Delay in onset of TrAb in painful condition. Control reaction time = 154±15ms Pain = 189±19ms	Mean VAS 7.2 (taken from graph). Superficial injection of lidocaine included. Variable superficial muscle alteration in response to pain.	2.19 (EMG onset TrAb)	5,8,10-12,14,22- 25
Hodges et al.,(2003a)	5 male and 2 female Age: 28.6±3.6 No known LBP. Single shoulder flexion (standing)	1.5ml 5% Saline Depth: 35mm Location: 60mm lateral to L4	Temporal and Spatial parameters of EMG (related to reaction times). Intramuscular EMG: deep LM, superficial LM, OE and OI.	Delay in onset of TrAb-actual difference (ms): Control= -18.9±2.4 Isotonic Saline = -13.7±4.0 Hypertonic Saline = -8.6±4.1 Early onset of DM-actual difference: Control=47.3±5.5 Isotonic Saline = 47.7±5.8 Hypertonic Saline = 24.4±5.5	Mean VAS 6.2±1.0 Superficial injection of lidocaine included. Significant difference only between control and 5% saline, not isotonic and hypertonic saline for TrAb.	3.31 (EMG onset TrAb) 4.49 (EMG onset deep LM)	5,8,10-12,14,22- 25
Moseley and Hodges,	7 male and 9 female Age: 24±5	Noxious cutaneous electric	Temporal and Spatial parameters of EMG (related to reaction	Significant and progressive delay in TrAb/OI and significant and progressively earlier onset	No actual figures presented. Resolution of pre-	NA	5,7,8,10- 12,14,22-25

(2005)	No known LBP. Single shoulder flexion (sitting)	stimulation placed bilaterally over the PSIS.	times). Surface EMG placement: TrAb/OI, EO and deltoid.	of OE during pain trials.	pain results on removal of noxious stimulus.		
Kiesel et al.,(2008)	6 male Age: 26±7.3 No known LBP. Abdominal 'drawing-in' maneuver (crook lying) and prone limb raising.	1.5ml 5% saline Depth: 35mm Location: 60mm lateral to L4	Ultrasound measured thickness of TrAb and LM at rest and during contraction.	Significant difference in thickness of TrAb and LM during contraction.	6 of seven reached required pain level of ≥4 (VAS) Optional 0.5ml 1% lidocaine subcutaneously to make deep injection more comfortable. Actual differences 0.68±0.08 v's 0.59±0.07 with pain present (50% change in thickness control and 29% change in pain condition. Differences with LM range 0.26-0.32cm.	1.31 (US thickness TrAb) 0.53-0.63 (US thickness LM)	5,8-12,14,22-26
Dickx et al.,(2008)	15 male Age: 23.33±0.82 No known LBP. (1) Supine MRI (2) Sorenson test (10 repetitions) then MRI (3) Saline, exercise then MRI	1.5ml 5% Saline Depth: 25mm Location: 40mm lateral to L4	Muscle functional MRI (measuring shifts in signal intensity caused by increased transverse relaxation time (T2) of muscle water in response to activity)	Significantly decreased muscle activity in LM and erector spinae during painful exercise. Results show difficulty activating in the presence of pain.	5.3-5.9 VAS during exercise. Exercised at 40% of 1RM	0.000032-0.58 (shifts in T2 value in ms)	5,8,11,12,14,15,22-25

EMG, electromyography; VAS, visual analogue scale; L1, 2, 3, 4, 5, respective lumbar vertebrae; Th12, 12<sup>th</sup> thoracic vertebrae; FRR, flexion relaxation response; AP, anterior-posterior; ML, medio-lateral; LM, lumbar multifidus; OE, obliquus externus; OI, obliquus internus; TrAb, transversus abdominus; US, ultrasound scan; MRI, magnetic resonance imaging; RM, repetition maximum.

Table 2.3. Experimental pain-relief studies.

Authors	Subjects and Tasks	Pain relief model	Outcome measures	Results	Comments	Effect size	Down and Black missing criteria
Rashiq et al.,(2003)	17 male and 11 female Age: 54, range 23-78. CLBP duration: 127, range 10-420 months Sorenson test	IV fentanyl 1µg/kg. Significant pain relief reported (0.9 on VRS)	Sorenson test time	Saline injection Sorenson sore = 60±42s Fentanyl injection Sorenson score = 77±49s	Minimal pain relief. Large variation in performance.	0.38	5, 8, 12, 25
Holm et al.,(2000)	38 male and 49 female Age: 48, range 22-79 CLBP duration: 12.3 yrs Mixed diagnosis including Spondyloysis, Spondylolithesis, post disc surgery, fracture sequelae. Isokinetic Dynamometer for trunk flexion/extension at 60°/s and 120° /s.	Intra-articular combined steroid and anaesthesia into L5/S1 ± L4/5 bilateral ZA joints.  Significant pain relief (21.2mm).	Total work (joule) at 60°/s and 120°/s.	No significant difference in performance following injection for 60 or 120° /s.	Weak correlation between decrease in pain and increase in muscle performance.	0.0003-0.18	3, 5, 8-12, 14, 15, 21, 23-25
Lilius et al.,(1989)	48 male and 61 female Age: 44, range 19-64	Intra-articular steroid injection to ZA joints. Pericapsular steroid	VAS (pain) Disability score. ROM. All taken 1 hour	Significant improvements in likert scale for disability. Improvements in: Flex pre-23.0±17.6cm	ROM measured with tape measure. Rotation measured with a compass.	0.08	5, 11, 12, 21, 22, 24, 25

	CLBP >3/12 ROM and observed scored disability battery (blinded scorer)	injection. Intra-articular saline injection. Significant pain relief across groups (mean: 18.3mm).	upon completion of injection.	Flex post-24.4±18.8cm Rot pre = 47.8° Rot post = 49.3° No immediate differences with all other ROM.	Questionable clinically significant result.		
Jarzem et al.,(2005)	29 male and 21 female Age: 38.9±14.5 CLBP >3/12 (no leg pain) Flexion/Extension ROM Isometric deadlift Back extensions, sit-ups, side-flexions, oblique sit-ups	TENS (20 minutes duration) (50% reduction in pain) Varied electrode placement	VAS (pain) ROM (gravity goniometer) Isometric lifting capacity Maximum repetitions of sit-ups, back extensions, sideflexions, oblique sit-ups	Statistically significant gain in ROM. 15% gain in isometric lift capacity post 'real' TENS. Statistically significant increase in maximum repetitions for all tasks	Real numbers not reported.	NA	5,7,8,11,12,21, 22,24,25
Davis and Kotowski, (2005)	3 male and 3 female Age: 42.7±13.8 4 male and 2 female Age: 37.0±4.7 6 female Age: 44.3±8.5 2 male and 4 female 44.0±21.1	Acupuncture  Chiropractic  Massage  Physical Therapy	Questionnaire (Adapted RMDQ and NASS LSOAI) Functional capacity evaluation (LMM)	<b>Acupuncture</b> 4.8; 4%* increase lat vel; lat acc 8*; -7.2% increase/decrease in Tw vel; Tw acc <b>Chiropractic</b> 14.1; 15.9% increase in lat vel; lat acc 50.3; 50.6% increase in Tw vel; Tw acc <b>Massage</b> 15*; 11%* increase in lat vel; lat acc 30*; 28%* increase in Tw vel; Tw acc	No direct measurement or reporting on pain scores.	NA	3-8, 11, 12, 14, 15, 21-25

				PT 12*; 7.5% increase in lat vel; lat acc 2.6; -2.3% increase/decrease in Tw vel; Tw acc			
Simmonds and Rebello,(2003)	5 male and 10 female Age: 44.4±6.4 LBP (undefined acute or chronic) Matched controls. Repeated sit to stand (5) at 3 self varied speeds (slower than usual, preferred and fastest).	Superficial heat wrap to both groups – 40mins. Significant pain relief (t=3.2, p=0.006, actual numbers not reported)	VAS (pain) Total time to complete sit to stand task.	Pain relief resulted in significant increase in preferred speed only.	Actual difference in region of 5s (calculated from <sup>36</sup> ).	NA	Conference proceeding only

IV, Intravenous; VRS, verbal rating scale; CLBP, chronic low back pain; ZA, Zygapophyseal; ROM, range of motion; VAS, visual analogue scale; Flex, Flexion; TENS, transcutaneous electrical nerve stimulation; RMDQ, Roland Morris Disability Questionnaire; NASS LSOAI, North American Spine Society Lumbar Spine Outcome Assessment Instrument; LMM, Lumbar Motion Monitor; \*, data retrieved from graph; lat vel, lateral velocity; lat acc, lateral acceleration; Tw vel, twisting velocity; Tw acc, twisting acceleration.

### **2.3 Need for the study**

This literature review has identified a need for the exploration of new motion analysis technologies to overcome the limitations highlighted in this section. There is a clinical need to be able to measure curvature in LBP sufferers within a clinical setting at a cost and complexity that is acceptable to clinicians, without constraints alluded to in this review. Furthermore there is a need to be able to resolve kinematics in LBP sufferers within the clinical setting including the capabilities of measuring higher order kinematics. However just observing movement abnormalities is insufficient as it has already been well documented that LBP displayed altered kinematics and muscle functions. It is therefore important to attempt to determine the mechanisms behind such changes. Clinicians are often faced with the management challenge of LBP sufferers and this can involve attempts to restore the biomechanical functions of the trunk. Clinicians are therefore faced with trying to determine how best to alter these biomechanical functions, which lies in hypothesising as to the underlying mechanism causing such changes. It is commonplace for this to involve targeting pain as a way to influence biomechanical functions. However it is yet to be determined whether targeting pain will lead to an alteration of trunk biomechanics. Therefore there is a need to explore new motion analysis technology as well as determine the effects of pain relief on lumbar kinematics to aid in the ability to measure and understand movement behaviour in LBP sufferers.

In order to address these needs, the primary elements of this work involve the following investigations:

1. The reliability and validity of a novel fibre-optic device for the dynamic measurement of lumbar curvature.
2. Lumbar curvature, including peak curvature and sequencing of curvature change, in acute and chronic LBP sufferers.

3. The effects of pain relief on curvature, including peak curvature and sequencing of curvature change, in acute and chronic LBP sufferers.
4. The reliability of using inertial sensors for motion analysis, including range of motion, velocity and acceleration, within a clinical setting, in acute and chronic LBP sufferers.
5. The feasibility and reliability of using spatial domain analysis to visualise movement trajectory and as a method to measure movement irregularity.
6. The effects of pain relief on lumbar kinematics, including range of motion, velocity, acceleration and movement irregularity, during spinal movement.
7. The effects of pain relief on the lumbar muscles of iliocostalis and multifidus, including pattern, muscle onset and peak activation.

## **Chapter 3:**

### **Lumbar curvature**



### **3.1 Dynamic measurement of lumbar curvature using fibre-optic sensors**

#### **3.1.1 Introduction**

The previous chapter has provided an extensive review of lumbar curvature measurement including the advantages and limitations of various methods. It is evident that there is a need to explore new methods of curvature measurement with the aim of providing a system that can resolve dynamic curvature, provide a continuous representation of spinal shape and have minimal or no environmental constraints.

Fibre optic sensors have been used in industry to measure dynamic changes in shape and it is likely that similar technologies could be applied to the lumbar spine. Light flow through a specifically adapted sensor will be modulated by the degree of bend and therefore light intensity can be used to calculate curvature change. If a string of sensors registering this change in light flow were spaced minimally apart then through a process of smoothing, the output can produce a curve along the entire ribbon of sensors. The feasibility of such a tracking method to monitor body movements has been demonstrated using video game animation, and pilot studies have shown small errors during measurements of simulated joint motion (Morin et al., 2002). This attractive tool has the potential of being used to dynamically record the curvature of the spine, but its reliability and accuracy need to be established before it can be recommended for research and clinical use.

#### **3.1.2 Aim of study**

The purpose of this investigation was to examine the reliability and feasibility of a new fibre-optic motion analysis system for the dynamic measurement of lumbar curvature and to compare its repeatability with a conventional motion analysis system based on video methods.

### 3.1.3 Methods

#### 3.1.3.1 Participants

Thirteen participants (11 male and 2 female, age  $25.9 \pm 2.6$ , mean height =  $1.74 \pm 0.14$  m, mean weight =  $76.64 \pm 15.3$  kg) were recruited from the University of Roehampton.

Participants had no history of low back pain in the previous twelve months and were excluded if they had any history of spinal surgery, tumours or disorders of the lower or upper limb that may be aggravated by the test procedures. All participants were clinically examined by a physiotherapist, and none of them was found to exhibit any spinal motion restriction or observable spinal deformities.

This study was approved by the Ethics Committee of the School of Life Sciences, Roehampton University. Written consent was obtained following explanation of experimental procedures and potential risks.

#### 3.1.3.2 Instrumentation

##### Fibre-optic system

A series of 8 paired fibre-optic sensors, attached to a ribbon of sprung steel (S128048NL 'Shapetape<sup>TM</sup>', Measurand, New Brunswick, Canada) (size = 480mm x 13mm x 1.3mm) was used to measure spinal motion and curvature. These sensors are specifically treated to measure curvature through the measurement of transmitted light intensity. A base reference sensor was attached securely to the skin overlying S1 with each sensor pair subsequently spaced every 60mm. Position (Cartesian coordinate) data is calculated by evaluating and averaging the total curvature change across each 60mm sensing region, producing a vector describing the spatial position of the end of the sensing region relative the beginning. This positional data for each sensor is then combined for the length of the spine resulting in a sagittal profile of spinal shape (figure 3.1.1). The fibre optic ribbon was fed into a modified

elastic bandage housing which would allow it to slide inside during movements of the spine (figure 3.1.1). Pilot testing showed this to be the most efficient attachment. The interface box to which the base sensor was attached was linked to a personal computer and positional data were captured using software developed by the company (Measurand Inc, New Brunswick, Canada) at 100Hz and stored for later processing.

#### Video system

A nine camera, synchronised, three dimensional optical motion analysis system was used (Vicon 370). The cameras were positioned in a ring setup and calibrated prior to all data collection. Reflective markers (diameter 13mm) were attached to the skin overlying the Anterior Superior Iliac Spines (ASIS) and Posterior Superior Iliac Spines (PSIS), along with eight markers passing superiorly from S1 at 60mm intervals (figure 3.1.1). A local coordinate system origin was set up using the pelvic markers and the coordinates of the spinal markers were transformed with respect to the local pelvic coordinate system and were fitted with a spline curve to provide a sagittal profile of spinal shape (figure 1). Motion was captured at 100Hz and stored for later analysis.

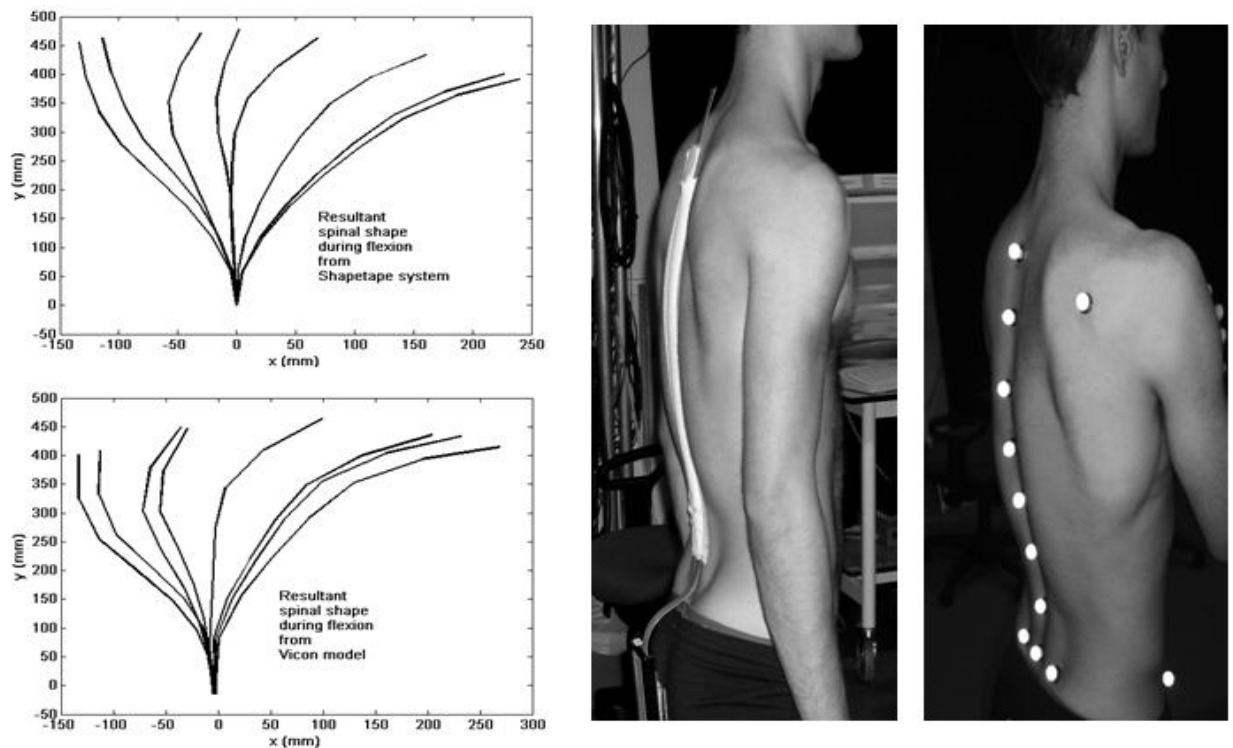


Figure 3.1.1. Fibre-optic and video system attachment and resultant sagittal curves during flexion.

### 3.1.3.3 Procedure

Participant's height and weight were recorded and the location of S1, L3 and L1 spinous processes were identified by the same experienced manual therapist. Along spine measurements were taken for the distance between S1 and L1, along with S1 and L3 in upright standing, full flexion and once in position about to lift an object (a crate which measured 310mm x 450mm x 450mm and weighed 2 kg), in order to determine the location of L1 or L3 relative to S1 at the extreme spinal postures. This enabled the spine to be separated into the whole lumbar spine and lower lumbar spine. S1-L1 was chosen as it represents the boundary of the lumbar spine and offers easy comparison to existing literature, while S1-L3 was chosen to determine if the system could reliably analyse small regions within the lumbar spine. Subjects were assigned randomly (coin toss) to either use the video system or fibre-optic system first. Subjects were instructed to stand bare foot on assigned markers and focus on a wall marker set at 1.5 meters high, with arms relaxed by their side. They were requested to

bend forwards as far as possible, pause for a second before returning upright. Identical instructions were given for bending to lift a crate. The crate was positioned using floor markers to ensure identical start positions and all movements were completed three times. The data acquired during the three trials were averaged. Subjects who tested the fibre-optic system first then had the fibre-optic system removed and proceeded to the video system. Once retesting was complete, the fibre-optic system was reattached and testing repeated. Subjects who were assigned to the video system first completed testing using the video system, then the fibre-optic system, which was removed and reattached for final testing. Therefore, in total, three distinct sets of data collection were completed; fibre-optic 1st, fibre-optic 2<sup>nd</sup> and video.

#### 3.1.3.4 Data Analysis

All raw data were analysed using Matlab (Mathworks, R2008b). The position data of the fibre-optic and video systems in the sagittal plane, which both refer to the local sacral coordinate system, were fitted with a piecewise cubic hermite interpolating polynomial (Matlab function `pchip`) and tangents to the curve were calculated using two consecutive data points at two defined landmarks - S1 and the desired lumbar level (L1 or L3). A piecewise cubic hermite interpolating polynomial is a third degree spline where each polynomial consists of two control points and two control tangents. It was chosen following a series of curve fitting trials as it was deemed to offer a more realistic representation of expected shape especially at the S1 region as compared to the more conventional cubic spline fit. Details of curve fitting trials can be found in appendix 4.0. Curvatures of the spine were derived from the angles between the two tangents. As motion occurs, the location of L1 (or L3) changes relative to S1, therefore curvature across time was adjusted, according the predicted location of the lumbar spinal process relative to S1. The prediction was achieved through the determination of the length change (measured difference between the S1-L1 (or L3) in standing and at the extreme of motion) and the determination of curvature change (difference between standing curvature and extreme curvature). A scaling factor was then calculated utilising the magnitude of

curvature change to determine the appropriate length change. Therefore adjustments to length were made relative to the degree of curvature change across time. The above adjustment enabled the calculation of the superior tangent to more accurately follow the L1 (or L3) segment during motion. Adjusted curvature-time data were calculated following the determination of motion onset and offset. Motion onsets and offsets were determined by first identifying a region of static standing, prior to motion to calculate the resting mean lumbar curvature and standard deviation. Motion onset was defined as the point when the curvature remained greater than the mean plus 3 times the standard deviation for 30 samples. Resultant curvature-time plots were time normalised in order to take into account the differences in speed at which the subjects performed the movement. The coefficient of multiple correlation (CMC) (Li & Caldwell, 1999a) and the root mean square (rms) error were calculated to determine the similarities between the three movement trials. The CMC value approaches 1 when the curves are similar and 0 if they are dissimilar. Repeated measures reliability for both the whole lumbar spine (S1-L1) and lower lumbar spine (S1-L3) for the fibre-optic system and video system were calculated across three repeated movement trials. Reattachment reliability was measured for both regions using the mean time normalised curvature-time curve for the first series of trials compared to the mean curve of the second series of movement trials collected by the fibre-optic system. Finally comparisons between the video and fibre-optic systems were calculated by first using a paired t-test to check for any difference between first and second data capture with the fibre-optic system. If no significant difference was detected, the data would then be combined to utilise the mean of all six trials for the fibre-optic system to compare with the mean of the three video system trials. Bland and Altman plots were used to compare the level of agreement for peak curvature.

### **3.1.4 Results**

The mean (sd) actual peak curvature values for the fibre-optic system were  $-11.5^{\circ}$  (7.3) and  $-7.4^{\circ}$  (6.6) for flexion and lifting of the whole lumbar spine, compared to  $-10.6^{\circ}$  (8.3) and  $-7.9^{\circ}$

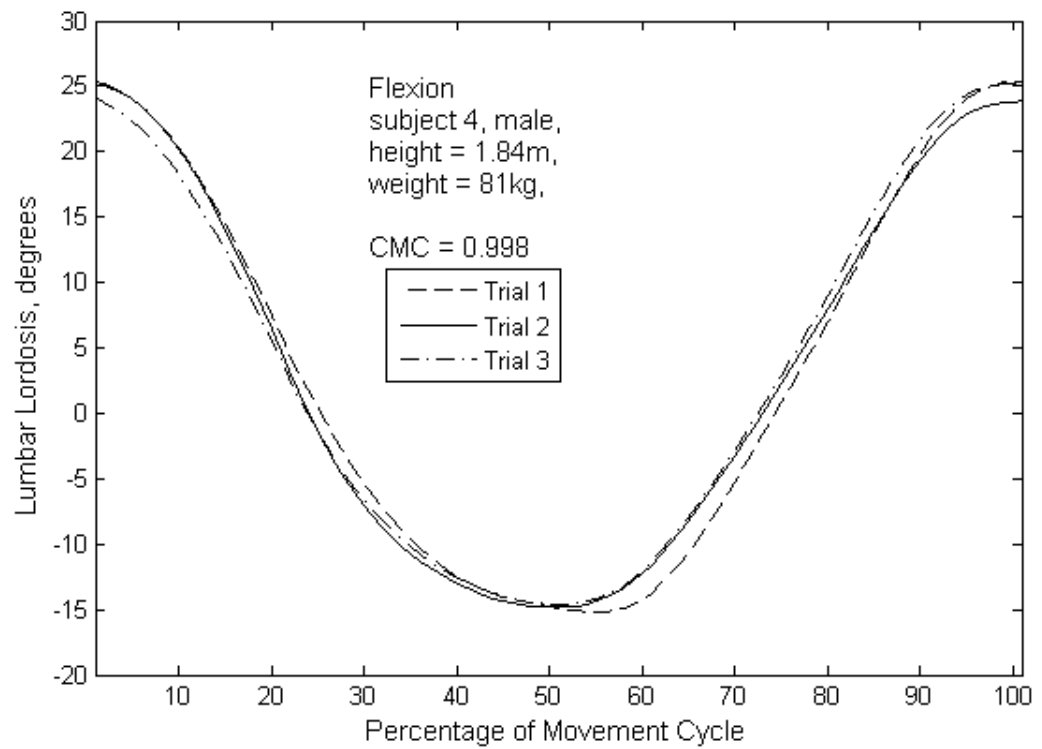
(7.8) for Vicon respectively. The lower lumbar spine mean (sd) values were  $-4.9^{\circ}$  (3.8) and  $-3.4^{\circ}$  (2.5) for flexion and lifting as measured with the fibre-optic system compared with  $-4.4^{\circ}$  (4.1) and  $-4.3^{\circ}$  (3.7) for Vicon respectively.

The mean CMC values were found to be excellent across both movements and regions of the lumbar spine (table 3.1.1).

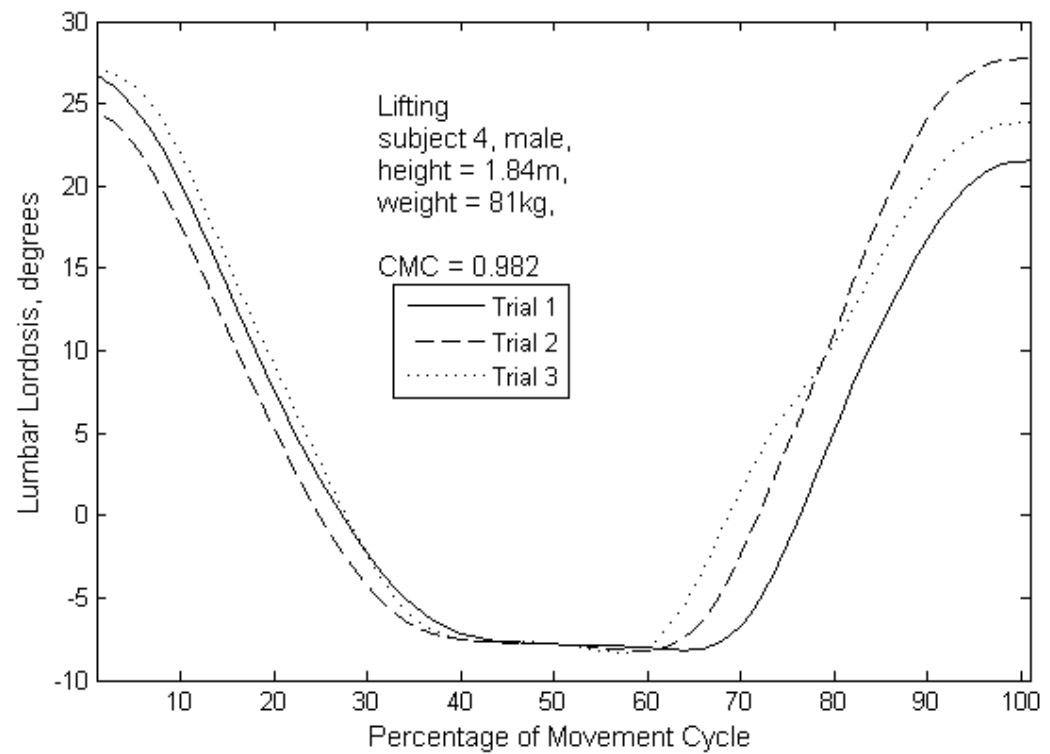
Table 3.1.1. The mean (sd) CMC for repeated measures of the whole lumbar spine and lower lumbar spine.

<b>Whole Lumbar Spine (S1-L1)</b>	<b>Flexion</b>	<b>Lifting</b>
Fibre-optic r	0.98 (0.02)	0.98 (0.02)
Video r	0.97 (0.02)	0.97 (0.04)
<b>Lower Lumbar Spine (S1-L3)</b>	<b>Flexion</b>	<b>Lifting</b>
Fibre-optic r	0.97 (0.03)	0.97 (0.03)
Video r	0.93 (0.11)	0.96 (0.03)

The fibre-optic system offered almost identical consistency across the repeated measures testing for all regions and tasks compared with the video based system. The fibre-optic system shows excellent similarity for repeated measures (figure 3.1.2) and small variability as displayed by figure 3.1.3 showing resultant mean curves and 95% confidence bands for repeated measures testing. The confidence bands show the variability in movement patterns. The root mean square error magnitudes for each movement were found to be small across both systems (figure 3.1.4).



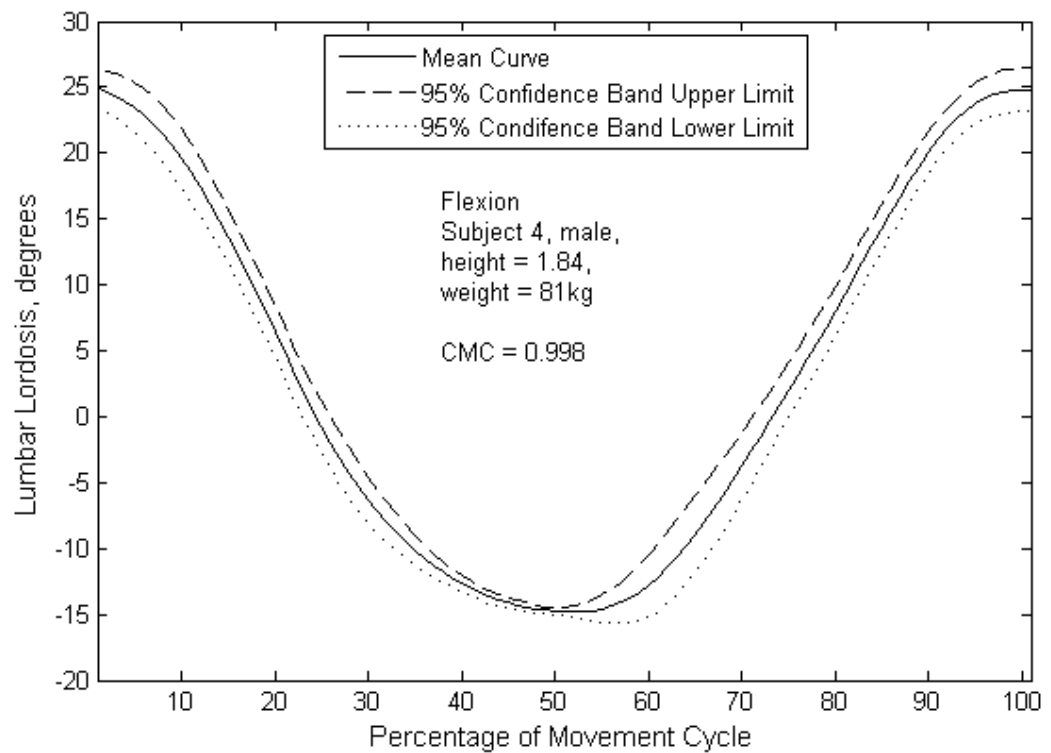
(a)



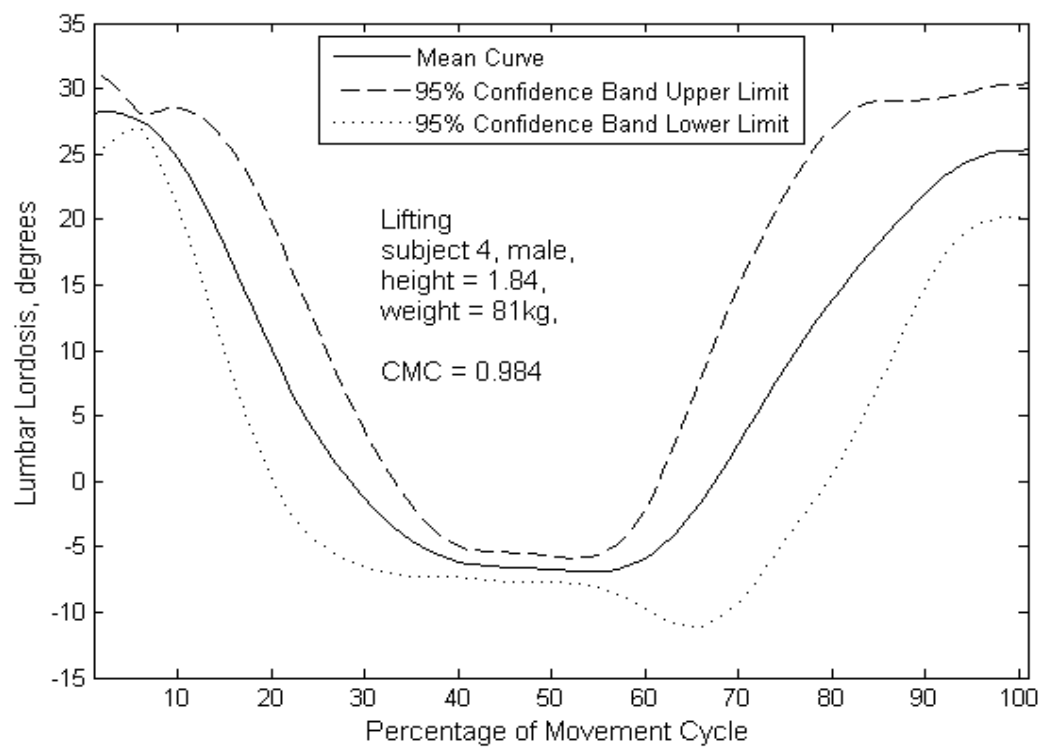
(b)

Figure 3.1.2. The normalised curvature-time curves of one participant for repeated measures reliability of (a) flexion and (b) lifting as recorded by the fibre-optic system.





(a)



(b)

Fig. 3.1.3. Mean normalised curvature-time curves with 95% confidence bands for (a) flexion and (b) lifting as recorded by the fibre-optic system.

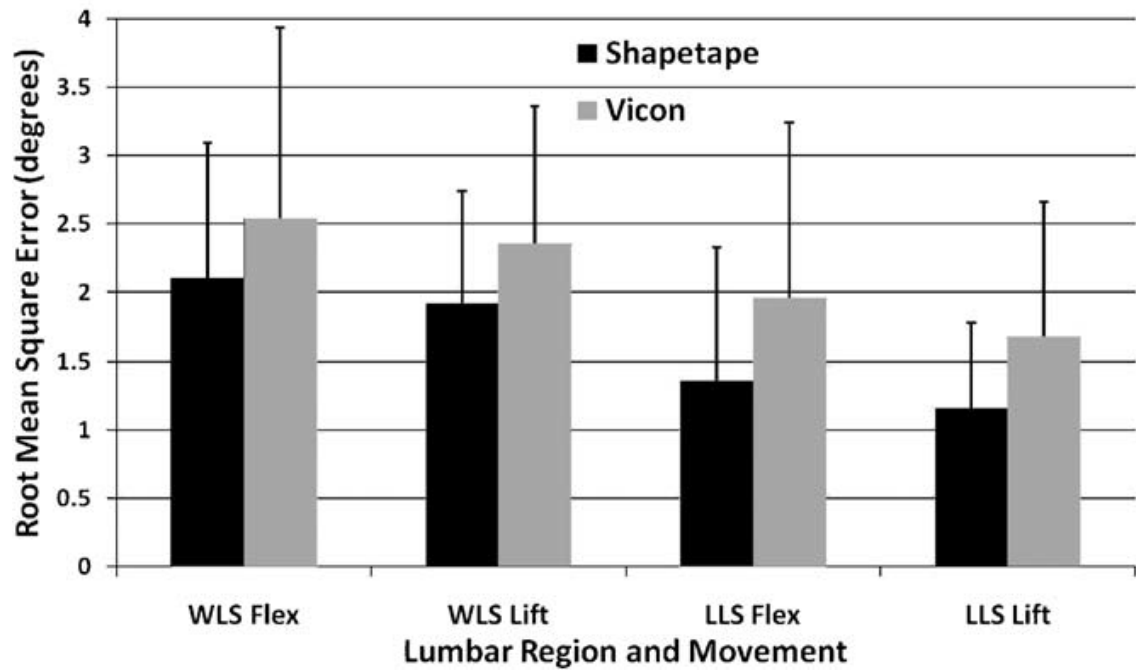


Fig. 3.1.4. The mean root mean square (RMS) error associated with measuring curvature with both systems. WLS, whole lumbar spine (L1-S1); LLS, lower lumbar spine (L3-S1). Vertical bars represent one standard deviation.

Reattachment reliability for the fibre-optic system was excellent for both regions and tasks (table 3.1.2). There were no-significant differences between the first and second data capture with the fibre-optic system ( $p = 0.945$ ).

Table 3.1.2. The mean (sd) CMC and root mean square (rms) error values for comparisons having removed and reattached the fibre-optic system.

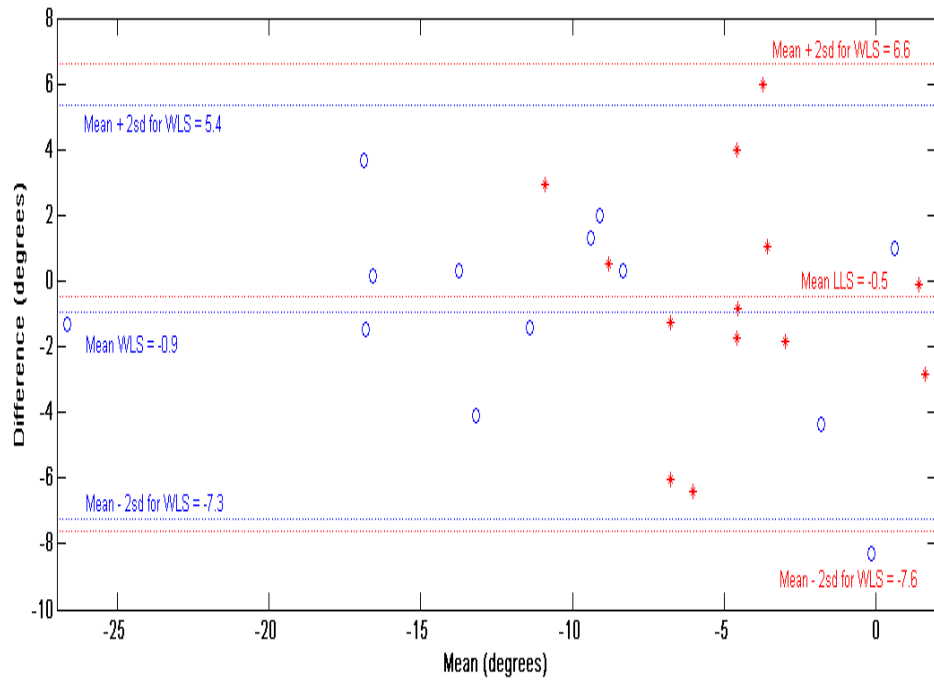
Whole Lumbar Spine (S1-L1)	Flexion	Lifting
r	0.97 (0.03)	0.97 (0.05)
rms (deg)	1.9 (1.0)	2.1 (2.0)
Lower Lumbar Spine (S1-L3)		
r	0.94 (0.08)	0.91 (0.14)
rms (deg)	1.6 (1.2)	1.9 (1.8)

There was excellent agreement, as noted by high mean CMC and low rms error values between the fibre-optic and video based systems for the whole lumbar spine and good agreement for the lower lumbar spine (table 3.1.3).

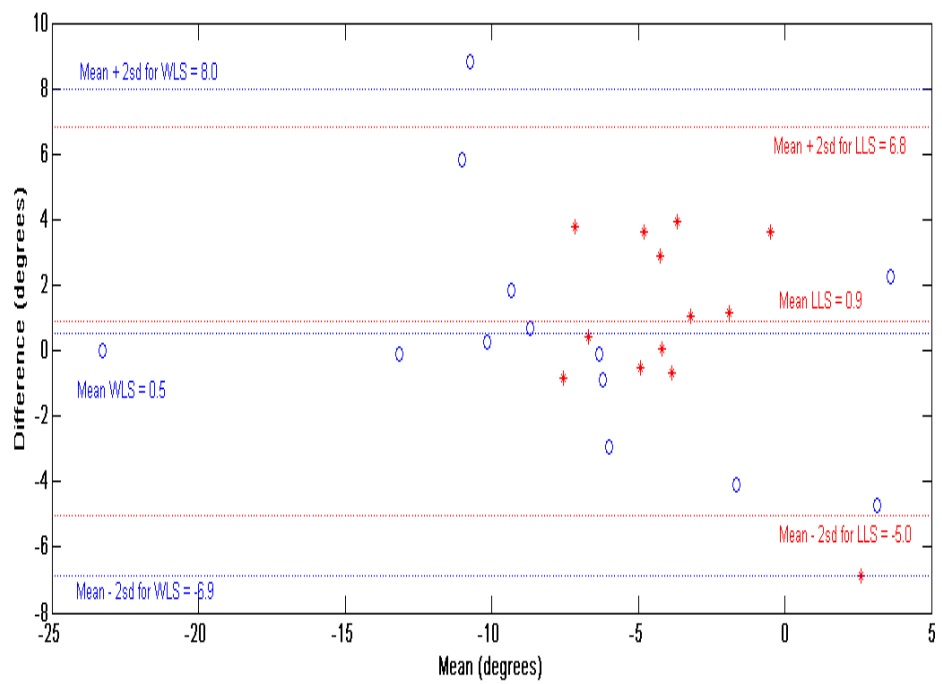
Table 3.1.3. The mean (sd) CMC and rms error values for comparisons of mean fibre-optic measured curvature with mean video measured curvatures.

<b>Whole Lumbar Spine (S1-L1)</b>	<b>Flexion</b>	<b>Lifting</b>
r	0.95 (0.04)	0.94 (0.06)
rms (deg)	3.5 (1.8)	3.4 (2.0)
<b>Lower Lumbar Spine(S1-L3)</b>		
r	0.86 (0.09)	0.84 (0.11)
rms (deg)	3.2 (1.9)	3.2 (1.4)

Good agreement between the two systems was observed for peak curvature measurements, as can be seen in the Bland and Altman plots of figure 3.1.5. The accuracy of the fibre-optic system as measured by the absolute mean difference for the peak curvature measured by the two systems for the whole lumbar spine was  $2.3 \pm 2.3^{\circ}$  and  $2.5 \pm 2.7^{\circ}$  and for the lower lumbar spine was  $2.7 \pm 2.2^{\circ}$  and  $2.3 \pm 2.0^{\circ}$  for flexion and lifting respectively.



(a)



(b)

Fig. 3.1.5. The difference on peak curvature obtained by the fibre-optic and video system against the mean for the specific movement.  $\circ$  = whole lumbar spine;  $*$  = lower lumbar spine, (a) flexion and (b) lifting.

### 3.1.5 Discussion

The results of study one indicate that the fibre-optic system is reliable for measuring dynamic lumbar spinal curvature, with the ability to resolve spinal shape change across time at multiple spinal regions. The movement-time behaviour was highly consistent for all tasks and regions, in fact more consistent than a matched video based method. The fibre-optic system can be reliably removed and replaced with data following the removal closely matching that previously collected.

The fibre-optic system exhibited similar repeatability when compared to the video system, for both regions, for the movements of flexion and lifting. The CMC values approaching 1 for the spinal curvature measurements show excellent agreement between the two systems.

Furthermore the minimal difference for peak curvature measurement shows that the fibre-optic system is accurate as compared to the video-based system. Importantly this similarity was observed even though the data collections were not simultaneous which introduces a level of biological error as it is very difficult to move identically for each trial and may explain the small inconsistencies and errors seen between the systems.

In this study the fibre-optic system was shown to be able to successfully measure changes in spinal curvature. The average values of peak curvature are compared to those of previous studies (table 3.1.4). One interesting finding was that small values of flexion curvature were achieved at the end of the flexion range. This suggests that flexion is primarily an unfolding of the lordotic curve with very little actual kyphosis achieved.

Table 3.1.4. Comparison of the peak curvature measurements of the present study with those in the literature.

Study	Methods	Region	Standing curvature (°)	Flexion peak curvature (°)
Present	Fibre-optic	L1-S1	25	-12°
Present	Opto-electronic	L1-S1	29	-12°
Thoumie et al., (1998)	Electro-goniometer	S1 – 10cm superior	38	-22°
Thoumie et al., (1998)	X-ray	S1 – 10cm superior	59	5
Mannion et al., (2004)	Spinal mouse	T12-S1	32	-33°
Kellis et al., (2008)	Spinal mouse	T12-L1	19-24	-33-42°
Dolan and Adams, (1993)	Isotrak	L1-S1	33-35	-22-26°
Ng et al., (2001)	Inclinometer	T12/L1-L5/S1	24	-28°
Youdas et al., (2000)	Flexicurve	T12-S2	39-56	-21-29°
Burdett et al., (1986)	Goniometer	T12/L1-S1	17	-16°
Burdett et al., (1986)	X-ray	T12-S1	50	-4°

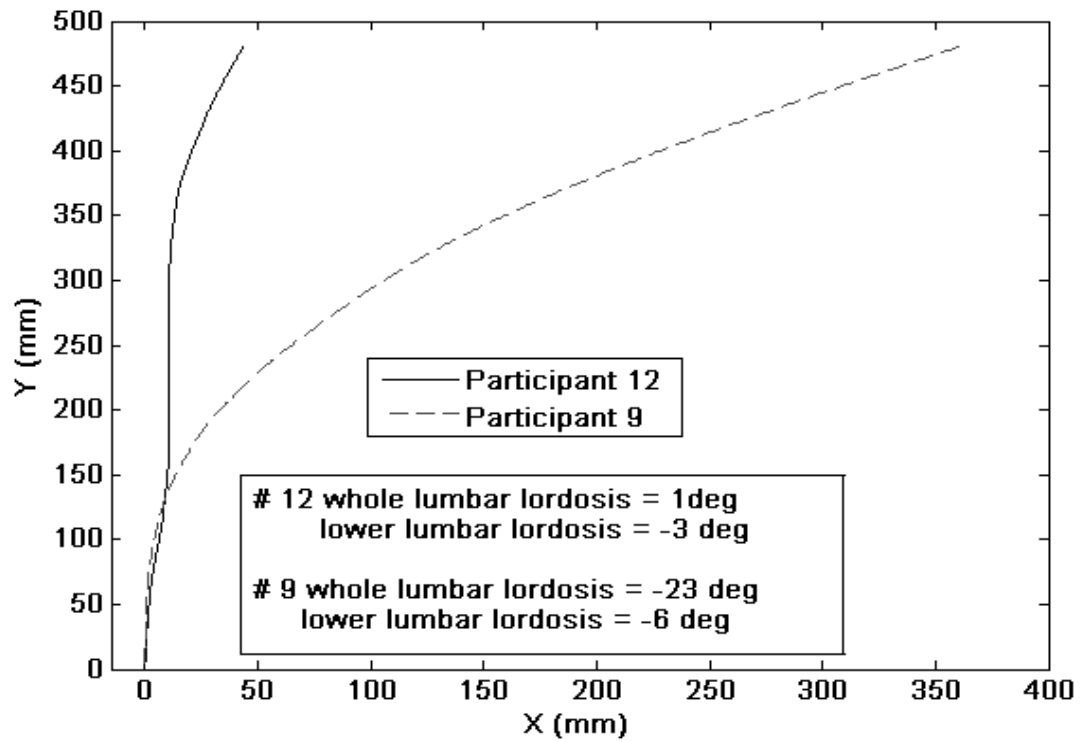
The clinical implication of this finding is that during forward bending, a significant amount of flexion comes from the hips and pelvis; not only from the spine. The lack of reversal of lordosis during forward flexion has also been reported in previous radiographic studies which provide useful information about the curvature formed by the underlying vertebrae (Burdett et al., 1986; Thoumie et al., 1998). However, previous studies which used skin mounted sensors or inclinometers (Dolan et al., 1993; Kellis et al., 2008; Mannion et al., 2004; Ng et al., 2001; Thoumie et al., 1998; Youdas et al., 2000) tended to report a larger amount of kyphosis during forward bending (table 3.1.4). This may be explained by skin movement artefact or failing to exclude sacral motion from that of the lumbar spine, which would then lead to overestimation

of lumbar curvature. The differences in values may also be due to different participant characteristics which would result in highly variable curvature measurements. Differing measurement protocols and verbal instructions could affect the degree of curvature achieved as would any physical constraints such as those used by Ng et al., (2001). Furthermore differing start positions and definitions of the boundaries of the lumbar spine may further explain some of the differences (Thoumie et al., 1998; Youdas et al., 2000). It is evident from the comparison that the overall range of motion measured by the fibre-optic device is small compared with that of the other methods. It is possible that the method of attachment and inherent stiffness of the device cause a reduction in achievable range of motion, however the range is similar to that of the video method suggesting perhaps the limited range of motion is due to participant characteristics not the measurement device.

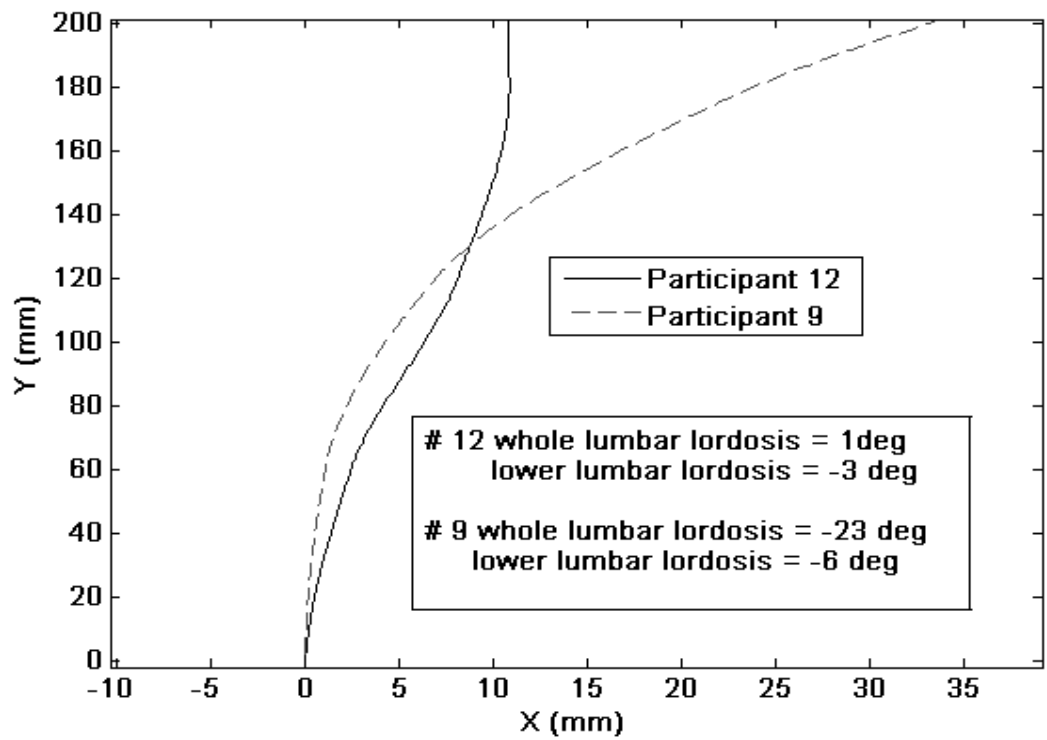
One attraction of the fibre-optic system is that it is simple to use and relatively cheap especially when compared to the video system used in this study. The cost of the fibre-optic system used in the present study is in the region of US\$4500-5000, but the exact cost will depend on the desired length and sensor configuration. This is certainly much less expensive compared to conventional video systems. The operator does not require more than a few days to become competent in the operation of the fibre-optic system. Data processing can be achieved using the operating software, or customised software which requires no training and can be used by an average clinician. It is portable, and not affected by the presence of metals, a common complaint with electromagnetic systems (Milne et al., 1996; Ng et al., 2009) or constrained by line of sight, as in video based systems. However the real benefits lie in the system's ability to measure dynamic change in spinal curvature, providing the investigator with the ability to study the curvature-time relationship. As LBP sufferers display altered spinal kinematics, especially temporal kinematics the ability to monitor dynamic movement behaviour is essential to observe such characteristics (Marras et al., 1999). Since the fibre optic system provides information along the whole length of the lumbar spine, information

regarding any region of interest can be obtained. This may be particularly important as sequencing movement behaviour of the spine is highly variable (Gatton et al., 1999) and clinical populations may adopt differing curvatures at different regions of their lumbar spine (Dankaerts et al., 2006). It may therefore be possible to use the device to study the sequencing of curvature change. This could be achieved by sectioning the curvature-time data into quartiles of time and calculating the degree of curvature change achieved during each quartile. Furthermore this could be measureable for small regional segments of the spine. This resolution of curvature across time for any region of the lumbar spine improves the flexibility and adaptability, increasing the potential application of such a system. It enables clinicians and researchers to observe sequencing behaviour of lumbar movement and visualise regional differences in posture and motion in the sagittal plane. This is highlighted by figure 3.1.6 which compares two participants' sagittal spinal shape at the moment of lifting. The significant difference in lordosis angle for the whole lumbar spine suggests different movement strategies are being utilised by these individuals (1 degree compared to -23 degrees curvature). However, this information alone fails to take in to account the overall spinal shape. It is evident that subject 12 has significant lower lumbar flexion during lifting, of a magnitude similar to subject 9 (-3 compared to -6 degrees). This suggests that the lower lumbar movement strategies are similar across these individuals and that significant regional differences in movement patterns are evident in the lumbar spine. This additional information only becomes available through the visualisation of continuous spinal shape, which can be offered by the new fibre-optic system. Clinically this is important as the degree of curvature influences load sharing across the motion segment (Adams et al., 1980a; Adams et al., 2000) and also affects the line of action of spinal muscles (McGill et al., 2000). This would be of great interest to not only the clinician studying a painful movement, or the surgeon reviewing the effect of surgery on spine kinematics but also in sports where the interactions between spinal curvature and performance could be analysed.





(a)



(b)

Figure 3.1.6. Sagittal view of two participants during lifting, (a) whole spine view, (b) close-up.

Moreover traditional systems use two sensors fixed to the skin overlying the two spinous processes, commonly whilst standing, which due to skin movement fails to closely represent the location of the same spinous process at the peak of the movement. The current system due to its attachment at S1 only and resultant continuous representation of spinal shape enable the location of the lumbar spinous process to be more accurately followed during motion.

The system can be operated up to 110 Hz, fast enough for the majority of kinematics applications and the system running with the operating software can provide real-time feedback which could aid the user in monitoring and modifying technique and developing position awareness of curvature during dynamic movement.

An important limitation is that back surface curvature measurement, as used in this and other kinematic studies, does not directly measure spinal curvature, therefore change in back surface curvature may not directly reflect changes in underlying spinal curvature.

Interpretation of the current findings may be limited by the sample size and to the tasks investigated. Participants were young and healthy with BMI values below 30.8; therefore the results may not be completely applicable to elderly or obese subjects. Further investigation in these populations would be useful. Limitations of the system relate to the sensitivity of the system to the base sensor attachment. This is because the fibre-optic system calculates the position and orientation of its sensors relative to this reference. Small sagittal tilts in the orientation of the base sensor will be registered as movement of the ribbon. For this reason applications should give extra caution to attaching the base sensor correctly which can be time-consuming. A limitation of the present method is that it did not measure lateral bending or twisting of the trunk, and the present system may not be useful for patients with idiopathic or sciatic scoliosis. The problem of the present system is that it comprises a flat ribbon which is stiff in bending sideways and cannot follow the shape of the spine if twisted. It is possible that

a more flexible fibre optic wire be developed in the future. It is also acknowledged that the sensitivity of the fibre optic device to a wide range of radii has not been established. It is possible that the accuracy of the measurement may be affected by smaller radii which may affect the recording of movements associated with large curvatures.

It is suggested that future research could also determine the impact of specific variables on dynamic curvature, such as pathology or LBP and how this affects regional sagittal kinematic behaviour which would further the understanding of the function of the spine and be important in the rehabilitation of spinal pain disorders.

### **3.1.6 Conclusion**

In conclusion, the fibre-optic system described in this study is able to provide highly reliable data regarding lumbar curvature. It spans the whole length of the lumbar spine enabling simultaneous capture of multiple regions of interest and can resolve curvature measurement through time. It is simple to use, relatively inexpensive and is capable of providing real-time feedback up to 110 Hz. Importantly, it does not require a designated operating zone or laboratory and therefore can be used anywhere, such as a clinic or the participants home.

## **3.2 Curvature measurement in LBP sufferers**

### **3.2.1 Introduction**

The previous chapter has demonstrated that it is possible to measure lumbar curvature using a novel fibre-optic device. It was shown to provide reliable and valid measurements of lumbar curvature as compared to a video based opto-electronic system (Vicon). The fibre-optic ribbon has the advantages of being fully portable, not affected by the operating environment and able to provide a continuous representation of spinal shape. However before it can be recommended for clinical use the functionality of such technology in LBP sufferers needs to be determined.

It has become common place for clinicians to measure lumbar curvature in the assessment of the lumbar spine (Dankaerts et al., 2006; Dankaerts et al., 2009; Scannell et al., 2003; Youdas et al., 2000; Youdas et al., 2006). Curvature has been linked to LBP (Dankaerts et al., 2006), has been shown to influence load sharing across the motion segment (Adams et al., 1980a; Adams et al., 2000; Wilke et al., 1999), to affect load tolerance of specific anatomical structures (Gallagher et al., 2005) and to alter the line of action of the trunk muscles (McGill et al., 2000). Dynamic sequencing of curvature change has been recommended in the assessment of LBP (O'Sullivan, 2000) despite not being studied in great depth in LBP sufferers. Previous studies have reported varied patterns of lumbar sequencing in individuals without LBP (Gatton et al., 1999; Kanayama, Abumi, Kaneda, Tadano, & Ukai, 1996), however this is yet to be studied in LBP sufferers.

### **3.2.2 Aim of study**

The purpose of this study was (1) to determine the reliability of the fibre-optic method to dynamically measure lumbar curvature in LBP sufferers and (2) to examine the lumbar

curvature, including peak curvature and sequencing of curvature change, in acute and chronic low back pain sufferers.

### 3.2.3 Methods

#### 3.2.3.1 Participants

Forty low back pain sufferers (20 acute and 20 chronic) were recruited from GP referrals and therapy departments and were routinely screened by a physiotherapist for inclusion and exclusion criteria (table 3.2.1). Participants were asked to rate their average pain over the preceding seven days using a visual analogue scale and complete a Tampa scale of kinesiophobia questionnaire (Vlaeyen, Kole-Snijders, Boeren, & van Eek, 1995). Participant demographics are displayed in table 3.2.2.

The study was approved by the National Research Ethics Service of the National Health Service, UK (reference number 08/H1111/38) and participants provided informed consent prior to participating.

Table 3.2.1. Inclusion and exclusion criteria

Inclusion	Exclusion
Pain confined to between lower ribs and inferior gluteal folds	History of tumors
Movement evoked pain	Spinal fractures
Aged 18-55 years old	Surgery
Seeking healthcare for LBP	Neurological signs or symptoms
Acute – Pain present for less than 3 weeks on a history of no pain for at least 12 months	Rheumatological or Neurological disease
Chronic – Pain present on at least 3 days per week for at least 52 weeks	Known spinal deformity

Table 3.2.2. Demographic characteristics of the participants.

	ALBP Mean (SD)	CLBP Mean (SD)	P-value
Male/Female	11/9	11/9	
Age (years)	42.7 (6.8)	36.6 (10.8)	0.0754
Height (m)	172.9 (11.3)	173.6 (11.2)	0.6994
Weight (kg)	82.6 (16.6)	83.7 (16.1)	0.5467
BMI (Kg/m <sup>2</sup> )	27.5 (4.0)	26.2 (4.1)	0.4337
TSK	39.0 (4.8)	38.9 (6.9)	0.8541
VAS (mm)	62.2 (16.6)	46 (22)	0.018
Duration	12.3 (6.7) days	9.4 (7.4) years	<0.0001

### 3.2.3.2 Procedure

Participant's height and weight were recorded and the location of S1, L3 and L1 spinous processes were identified by the same experienced manual therapist. Along spine measurements were taken for the distance between S1 and L1, along with S1 and L3 in upright standing, full flexion, full extension and once in position about to lift an object (a crate which measured 310mm x 450mm x 450mm and weighed 3 kg), in order to determine the location of L1 and L3 relative to S1 at the extreme spinal postures. This enabled the spine to be separated into the whole lumbar spine and lower lumbar spine.

Participants were instructed to stand bare foot on assigned markers and focus on a wall marker set at 1.5 meters high, with arms relaxed by their side. They were requested to bend forwards as far as possible, pause for a second before returning upright. Identical instructions were given for backward bending and lifting. The crate was positioned using floor markers to ensure identical start positions and all movements were completed three times. The level of movement evoked pain was recorded following the completion of the three movements, using a VAS where individuals recorded the worst pain experienced during that particular movement.

### 3.2.3.3 Data Processing

Data were processed using methods described in detail elsewhere (Williams, Haq, & Lee, 2010a). Briefly, raw sagittal plane data were spline fitted and curvatures derived from the

intersections of two tangents located at the S1 level and the L3 level for the lower lumbar spine and L1 level for the whole lumbar spine. Motion onsets and offsets were derived from an automated algorithm and data trimmed to remove the static portion. Curvature-time plots were time-normalised to take into account the different speed the participants performed the movement.

#### 3.2.3.4 Data analysis

The coefficient of multiple correlation (CMC) (Li et al., 1999a) and the root mean square error (RMSE) error were calculated to determine the similarities between the three movement trials. The means and standard deviations of the peak curvatures were determined and the repeatability of the peak curvatures were established using intra class correlations,  $ICC_{3,1}$  moreover mean absolute differences between peak measurements were also calculated. Repeated measures reliability for both the whole lumbar spine (S1-L1) and lower lumbar spine (S1-L3) were calculated across three repeated movement trials.

Mean peak curvatures for the whole and lower lumbar spine were calculated from the three movement trials and were compared across groups using independent t-tests (or Mann-Whitney U tests where normality could not be assumed).

Sequencing behaviour was calculated by indentifying the points of motion onset and peak curvature, which represents the motion from standing to end of range. This was then subdivided into quartiles according to time and the curvature change across each quartile calculated. The quartile of greatest change was determined and the number of individuals within each quartile of greatest change was calculated as a percentage. The frequencies of quartiles of greatest change were compared across ALBP and CLBP groups using Chi-squared test.

### 3.2.4 Results

A single participant repeated measures lifting graph is presented in figure 3.2.1.

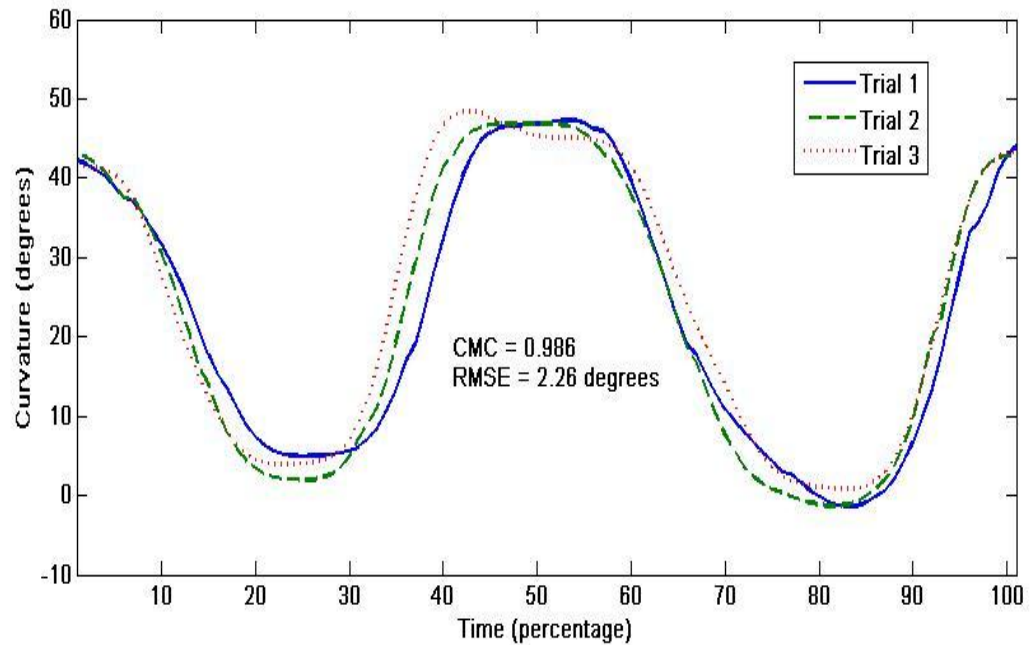


Fig. 3.2.1. Repeated measures graph for lifting of a single participant.

The mean CMC values were found to be excellent across all movements and regions of the lumbar spine with small variability (table 3.2.3). The root mean square error magnitudes for each movement were found to be small (table 3.2.3) with an overall mean RMSE of  $2.1^{\circ}$  (0.6) across all movements.

Table 3.2.3. Mean (sd) CMC and RMSE values for repeated measures reliability.

	ALBP – Whole Lumbar Spine (S1-L1)			CLBP – Whole Lumbar Spine (S1-L3)		
	Flexion	Lifting	Extension	Flexion	Lifting	Extension
CMC (r)	0.95 (0.02)	0.95 (0.05)	0.91 (0.07)	0.97 (0.02)	0.95 (0.04)	0.94 (0.05)
rmse ( $^{\circ}$ )	2.1 (0.9)	2.6 (1.2)	1.8 (0.8)	2.8 (1.3)	3.4 (1.3)	1.8 (0.8)
	ALBP – Lower Lumbar Spine (S1-L3)			CLBP – Lower Lumbar Spine (S1-L3)		
CMC (r)	0.95 (0.04)	0.95 (0.06)	0.81 (0.21)	0.97 (0.03)	0.95 (0.04)	0.86 (0.17)
rmse ( $^{\circ}$ )	1.5 (0.8)	1.8 (0.9)	1.4 (1.0)	2.0 (1.4)	2.4 (1.1)	1.4 (1.0)



The ICC values were excellent (0.99) for all repeated movement trials and absolute mean differences demonstrate small differences between repeated peak measurements (figure 3.2.2) with values  $<2.1^{\circ}$  for all movements.

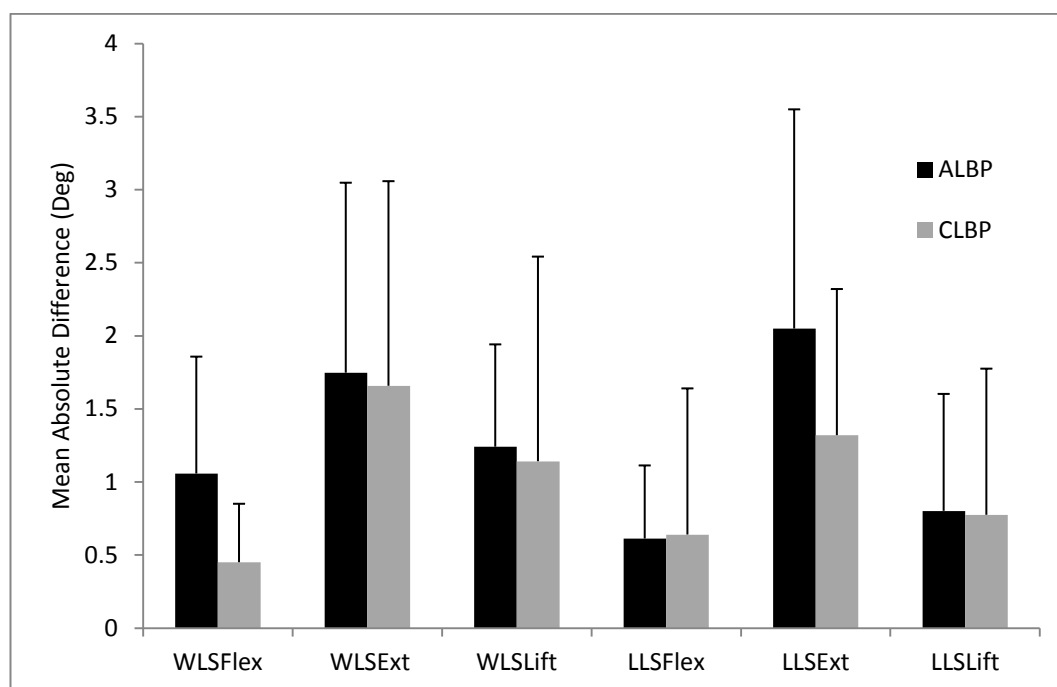


Fig. 3.2.2. Mean (sd) absolute difference between repeated peak curvature measurements (WLS, whole lumbar spine; LLS, lower lumbar spine; Flex, flexion; Ext, extension; Lift, lifting).

Peak curvature comparisons across the groups are presented in table 3.2.4. The mean (sd) level of evoked pain for the ALBP group was 41 (24), 32 (18), 41 (20) and for the CLBP group was 40 (20), 52 (25), 41 (24) for flexion, lifting and extension respectively as measured by the VAS.

The CLBP group displayed significantly greater lumbar curvature during flexion and lifting compared to the ALBP group for the whole lumbar spine (table 3.2.4 and fig. 3.2.3). Only lifting displayed significant curvature differences for the lower lumbar spine.

Table 3.2.4. Mean (sd) peak curvature comparisons across groups (degrees).

	ALBP	CLBP
<b>Whole Lumbar Spine (S1-L1)</b>		
Flex	<b>-7.0 (9.4)</b>	<b>-12.7 (6.3)*</b>
Lift	<b>-1.8 (11.7)</b>	<b>-10.2 (9.5)*</b>
Ext	44.1 (16.5)	47.9 (14.2)
<b>Lower Lumbar Spine (S1-L3)</b>		
Flex	-4.0 (6.1)	-6.1 (6.2)
Lift	<b>-0.5 (7.2)</b>	<b>-4.5 (6.9)*</b>
Ext	28.9 (13.3)	34.7 (20.1)

\*  $p < 0.05$ . Flex, Flexion; Ext, Extension; negative curvatures represent kyphotic curvatures; positive, lordotic curvatures.

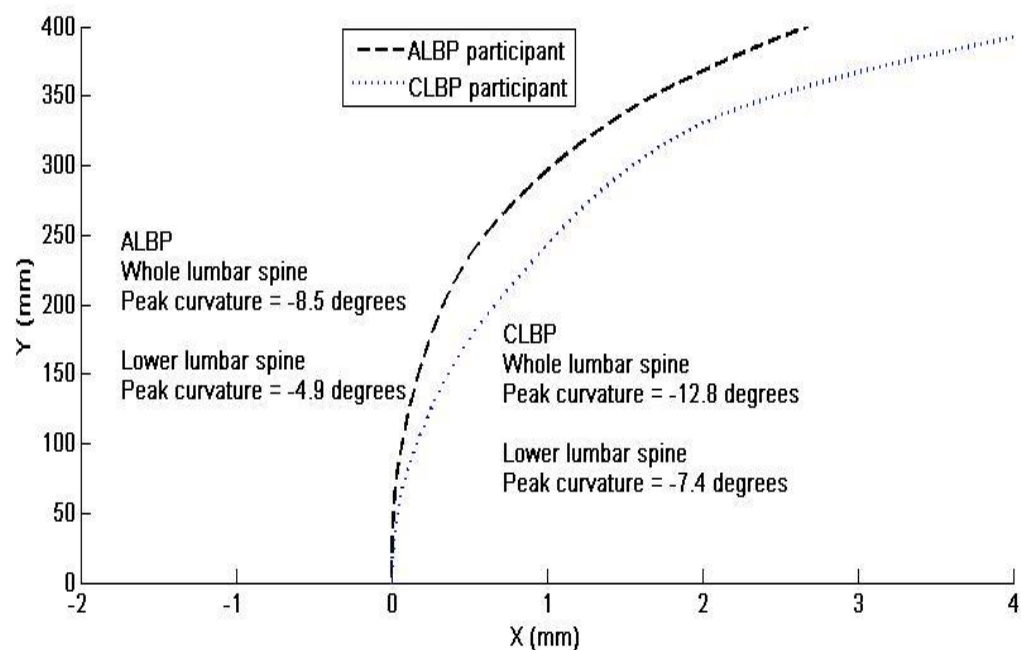


Fig. 3.2.3. Flexion peak curvature comparison between acute and chronic LBP.

An example of sequencing of curvature change for flexion is presented for a single participant in figure 3.2.4. Sequencing behaviour shows that both the ALBP and CLBP demonstrated the greatest curvature change in quartile 2 for flexion and lifting and quartile 1, and 2 for ALBP, for

extension of the whole lumbar spine (table 3.2.5). This finding was also mirrored by the lower lumbar spine with the exception of extension for the ALBP group. Chi-squared testing revealed that there was no between group differences in the frequency of specific quartiles of greatest curvature change ( $\chi^2 = 0.31-4.98$ ;  $p = 0.08-0.86$ ). Matching of the quartile of greatest change between the whole lumbar and lower lumbar spine occurred over 80% of the time for flexion and CLBP extension whereas it occurred on 47-65% of the time for the other movements.

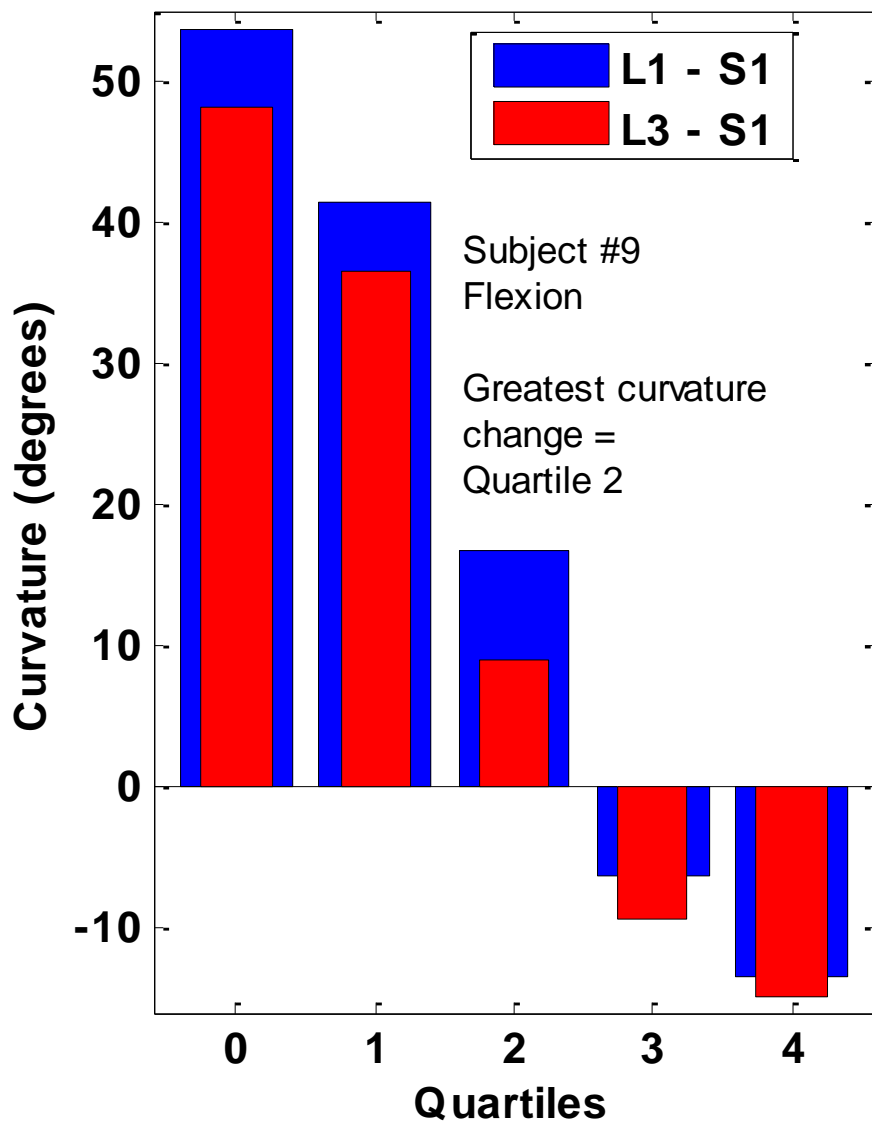


Fig.3.2.4. Sequencing of lumbar flexion for a single participant.

Table 3.2.5. Quartiles of greatest change presented as percentages.

	ALBP				L1 =	CLBP				L1 =
	Quartile of greatest change				L3 (%)	Quartile of greatest change				L3 (%)
	1st	2nd	3rd	4th		1st	2nd	3rd	4th	
Whole lumbar spine										
Flexion	30	70	0	0	80	10	74	16	0	89
Lifting	15	60	25	0	65	10	58	32	0	53
Extension	42	42	16	0	47	35	41	24	0	88
Lower lumbar spine										
Flexion	15	80	5	0	-	16	63	21	0	-
Lifting	10	65	35	0	-	0	68	32	0	-
Extension	47	26	21	5	-	41	29	29	0	-

L1 = L3, the whole lumbar spine quartile of greatest change matches the lower lumbar spine quartile of greatest change.

Correlations between pain and curvature were thoroughly explored but did not prove significant.

### 3.2.5 Discussion

This study presents a novel measurement method of dynamic lumbar curvature using fibre-optic sensors in LBP sufferers. The validity of the device has been previously reported by direct comparison to a video-based method in individuals without LBP, however it has now been shown to be reliable in the clinical setting with LBP sufferers. The system enables continuous data capture of multiple regions making this an attractive option for dynamic curvature measurement in LBP sufferers.

The results of this study demonstrate that ALBP sufferers achieve nearly 6 degrees less kyphosis of the lumbar spine during flexion when compared with CLBP. Greater kyphotic curvatures would increase tissue strain of the posterior spinal elements (Adams et al., 1980b;

Goel, Kong, & Han, 1993; McGill & Norman, 1986) and therefore this reduction in curvature may serve as a protective mechanism to avoid provocation of painful tissues. Previous studies have reported greater kyphotic curvatures using skin mounted devices than was measured in the current study (Burdett et al., 1986; Dolan et al., 1993; Kellis et al., 2008; Mannion et al., 2004; Ng et al., 2001; Thoumie et al., 1998; Youdas et al., 2000). These differences may be explained by skin movement artefact, differences in measurement protocols (Ng et al., 2001) or by variations in the boundaries of the lumbar spine (Thoumie et al., 1998; Youdas et al., 2000). However a previous study, with identical methods and technology, only utilising individuals without LBP, demonstrated a mean peak curvature of -12 degrees for flexion (Williams et al., 2010a). These results suggest that the CLBP sufferers in this study had no loss of curvature at the peak of flexion. This has been found in clinical trials where improvements in pain were noted with interventions targeting impairments other than curvature or ROM loss (Dankaerts, O'Sullivan, Burnett, & Straker, 2007; O'Sullivan, Twomey, & Allison, 1997). The clinical significance of such a finding might suggest that techniques designed to increase lumbar curvature may not be warranted for flexion related movement evoked pain in CLBP sufferers.

Peak curvatures achieved during lifting demonstrate that chronic LBP sufferers clearly adopt a more flexed lumbar spine during lifting, in the region of 8 degrees. This would result in significantly greater bending stresses on the lumbar spine (Dolan et al., 1993; Dolan, Earley, & Adams, 1994). In comparison to previous studies using individuals without LBP the mean peak lifting curvature was -7.4 degrees (Williams et al., 2010a). This magnitude of curvature was less than that observed in the CLBP group suggesting that CLBP sufferers adopt larger kyphotic curvatures than those without LBP. As mentioned previously this would load the posterior spinal structures more, potentially offering a mechanism of ongoing sensitisation.

Differences between ALBP and CLBP for extension curvature fail to show statistical significance which is likely due to the large standard deviation associated with this movement. These standard deviations suggest large inter-individual variation for this particular movement. Previous studies measuring extension curvature have shown mean values similar to the current study, ranging from 66° to 43° (Burdett et al., 1986; Ng et al., 2001; Thoumie et al., 1998). The actual measured differences between ALBP and CLBP groups were nearly 4 degrees for whole lumbar spine and nearly 6 degrees for the lower lumbar spine curvature. The magnitude of such a difference would be of clinical significance even though it did not reach statistical significance. Such a difference may suggest a loss of curvature in the ALBP which appears to be proportionally greater for the lower lumbar region. Therefore it may be evident that ALBP sufferers have difficulty creating lower lumbar curvatures during extension; however it is not yet known whether such an impairment represents a viable clinical target. The results of this study suggest that the sequencing behaviour of the lumbar spine for flexion and lifting is quite consistent with a large proportion of individuals displaying the greatest curvature change in the second quartile. This seems to be true for both acute and chronic LBP groups and for the whole and lower lumbar spines. The motion of extension displays greater variability with no fixed quartile of dominance. This is reflected in the actual curvature values which show very similar curvature changes across the quartiles.

Novel to this measurement system is the ability to compare the sequencing at the whole lumbar and lower lumbar region, which provides insights in the regional behaviour of the lumbar spine. Matching between whole lumbar and lower spines in sequencing behaviour was observed frequently in both groups. This may result in a coordinated curvature formation where the rate of curvature change is similar between these two regions of the spine. A deviation from this would result in either early lower lumbar kyphosis formation potentially increasing the strain on the lower lumbar spine, or a delay in the kyphosis of the lower lumbar spine. This delay would mean that the mass of the trunk would be shifted anteriorly prior to

the curvature change at the lower lumbar spine, increasing the load on this region whilst it is flexing. There has been some inconsistencies with the literature regarding sequencing as it has been previously reported that individuals complete forward flexion using a 'top down' movement sequence (Kanayama et al., 1996), which is in contrast to Gattton et al (1999) who found no such pattern instead reporting great variation in the order of segmental rotations. The current study aim was not to measure segmental motions or which segment moved first, moreover it focused on the quartile of greatest change and therefore the methodological differences may explain the differences in the results.

One limitation of this study is that the measurement was limited to the sagittal plane. The fibre-optic ribbon is flat and has inherent stiffness making its use for lateral bending limited and the adhesion method to enable the ribbon to slide during flexion fails to provide adequate attachment to measure rotation. The sample size was limited and constrained to those with LBP confined to the back therefore extrapolation to other types of LBP is not possible.

### **3.2.6 Conclusion**

The measurement of lumbar curvature is important in the evaluation of LBP. The fibre-optic method offers a reliable surface measurement tool for lumbar curvature and sequencing behaviour. This study demonstrated that acute LBP sufferers display reduced peak curvatures compared to chronic LBP for flexion and lifting. Sequencing of curvature change suggests that ALBP and CLBP sufferers create the greatest curvature change in the second quartile with flexion and lifting, and first with extension.

### **3.3 The effect of pain relief on lumbar curvature**

#### **3.3.1 Introduction**

The previous chapter has demonstrated that it is possible to reliably measure dynamic lumbar curvature in LBP sufferers, including peak curvature and sequencing behaviour. The previous chapter also found that ALBP sufferers achieve less curvature during flexion and lifting compared to CLBP. Sequencing demonstrated that the greatest curvature change was evident in quartile two for flexion and lifting and quartile one for extension in these groups of LBP sufferers. Despite these findings it remains unclear as to what is the underlying mechanism responsible for these movement patterns. It has been suggested that movement patterns in LBP sufferers represent an adaptation in response to pain (Shum et al., 2005a; 2005b; , 2007a), in an attempt avoid the provocation of pain, however to date this remains largely speculative. It is logical to assume that if this were the case then the elimination of pain should result in an automatic alteration in the lumbar curvature. The effect of pain relief on lumbar curvature has not been directly studied before, however the effects of pain relief on ROM has shown no or very small effects providing doubt to the above assumption (Jarzem et al., 2005; Lilius et al., 1989).

Despite the fact that the cause and effect relationship is not well understood, clinicians are faced with the challenge of clinical interpretation of lumbar curvature. It is possible that pain causes and alteration in curvature, however it is also possible that alteration in curvature causes pain. Management strategies often involve either targeting pain relief to alter the biomechanics of the trunk (Jette et al., 1994) or aim to influence curvature directly through the alteration of spinal posture and movement (O'Sullivan, 2000), however until the causative mechanisms are well understood these approaches remain speculative.



### 3.3.2 Aim of study

The aim of this study was to investigate the effects of pain relief on lumbar sagittal curvature during flexion, extension and lifting.

### 3.3.3 Methods

#### 3.3.3.1 Participants

Twenty acute and 20 chronic low back pain sufferers were recruited from therapy departments and GP referrals and were routinely screened by a physiotherapist for inclusion and exclusion criteria (table 3.3.1). Participants average pain in the week preceding data collection was rated on a visual analogue scale and a tampa scale of kinesiophobia questionnaire was completed (Vlaeyen et al., 1995). Participant demographics are displayed in table 3.3.2.

Table 3.3.1. Inclusion and exclusion criteria

Inclusion	Exclusion
Pain confined to between lower ribs and inferior gluteal folds	History of tumors
Movement evoked pain	Spinal fractures
Aged 18-55 years old	Surgery
Seeking healthcare for LBP	Neurological signs or symptoms
Acute – Pain present for less than 3 weeks on a history of no pain for at least 12 months	Rheumatological or Neurological disease
Chronic – Pain present on at least 3 days per week for at least 52 weeks	Known spinal deformity

Table 3.3.2. Demographic characteristics of the participants.

	ALBP	CLBP	P-value
	Mean (SD)	Mean (SD)	
Male/Female	11/9	11/9	
Age (years)	42.7 (6.8)	36.6 (10.8)	0.0754
Height (m)	172.9 (11.3)	173.6 (11.2)	0.6994
Weight (kg)	82.6 (16.6)	83.7 (16.1)	0.5467
BMI (Kg/m <sup>2</sup> )	27.5 (4.0)	26.2 (4.1)	0.4337
TSK	39.0 (4.8)	38.9 (6.9)	0.8541
VAS (mm)	62.2 (16.6)	46 (22)	0.018
Duration	12.3 (6.7) days	9.4 (7.4) years	<0.0001

TSK, Tampa scale of kinesiophobia; VAS, Visual analogue scale.

All participants provided informed consent and the study was approved by the National Research Ethics Service of the National Health Service, UK (reference number 08/H1111/38).

### 3.3.3.2 Procedure

Participant's height and weight were recorded and the location of S1, L3 and L1 spinous processes were identified by the same experienced manual therapist. Along spine measurements were taken for the distance between S1 and L1, along with S1 and L3 in upright standing, full flexion, full extension and once in position about to lift an object (a crate which measured 310mm x 450mm x 450mm and weighed 3 kg), in order to determine the location of L1 and L3 relative to S1 at the extreme spinal postures. This enabled the spine to be separated into the whole lumbar spine and lower lumbar spine.

Participants were instructed to stand bare foot on assigned markers and focus on a wall marker set at 1.5 meters high, with arms relaxed by their side. They were requested to bend forwards as far as possible, pause for a second before returning upright. Identical instructions were given for backward bending and lifting. The crate was positioned using floor markers to ensure identical start positions and all movements were completed three times.

The level of movement evoked pain was recorded following the completion of the three movements, using a VAS where individuals recorded the worst pain experienced during that particular movement. Participants were then requested to self-administer their usual oral analgesia and given a break of between 45-60 minutes after which the movements were repeated. There was no restriction on the type of analgesia used.

#### 3.3.3.3 Data Processing

Data were processed using methods described in detail elsewhere (Williams et al., 2010a). Briefly, raw sagittal plane data were spline fitted and curvatures derived from the intersections of two tangents located at the S1 level and the L3 level for the lower lumbar spine and L1 level for the whole lumbar spine. Motion onsets and offsets were derived from an automated algorithm and data trimmed to remove the static portion. Curvature-time plots were time-normalised to take into account the different speed the participants performed the movement.

#### 3.3.3.4 Data analysis

Peak curvatures for the whole and lower lumbar spine were calculated from the mean of the three trials and the effect of pain relief was determined using paired t-tests (or Wilcoxon signed-rank test when normality could not be assumed). Responders to pain-relief relief were determined as those who increased their kyphotic curvature for flexion and lifting by more than  $3^{\circ}$ , and lordotic curvature for extension, more than  $2^{\circ}$  following pain relief and Chi-squared testing was used to determine the frequency of response to pain-relief in each group.

Sequencing behaviour was calculated by indentifying the points of motion onset and peak curvature, which represents the motion from standing to end of range. This region was then sub-divided into quartiles and the curvature change across each quartile calculated. The quartile of greatest change was determined. Individuals with each quartile of greatest change

were calculated and converted into a percentage of the whole group. Responders were defined as those who shifted their quartile of greatest curvature change in response to pain relief and Chi-squared testing was used to determine the between group differences in the frequency of responders.

### 3.3.4 Results

A significant reduction in evoked pain was noted for each group and movement (table 3.3.3).

Table 3.3.3. The effects of pain relief on movement evoked pain.

	ALBP			CLBP		
	Pre VAS	Post VAS	p	Pre VAS	Post VAS	p
Flexion	44 (23)	21 (24)	0.005	39 (18)	9 (18)	0.004
Lifting	35 (19)	6 (13)	0.000	64 (19)	36 (27)	0.001
Extension	43 (17)	16 (20)	0.000	52 (22)	24 (24)	0.000

Pain relief medication choice is displayed in fig 3.3.1.

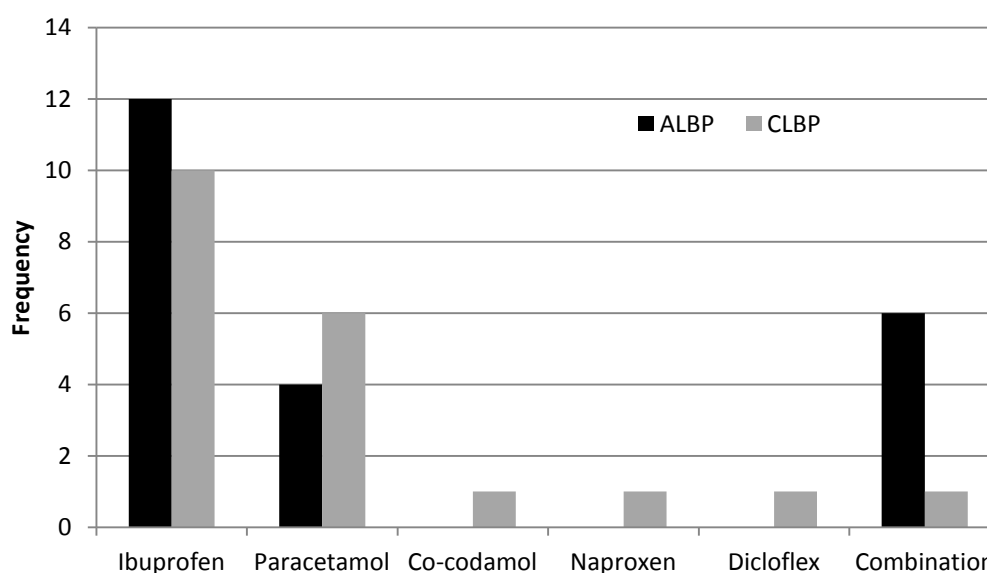
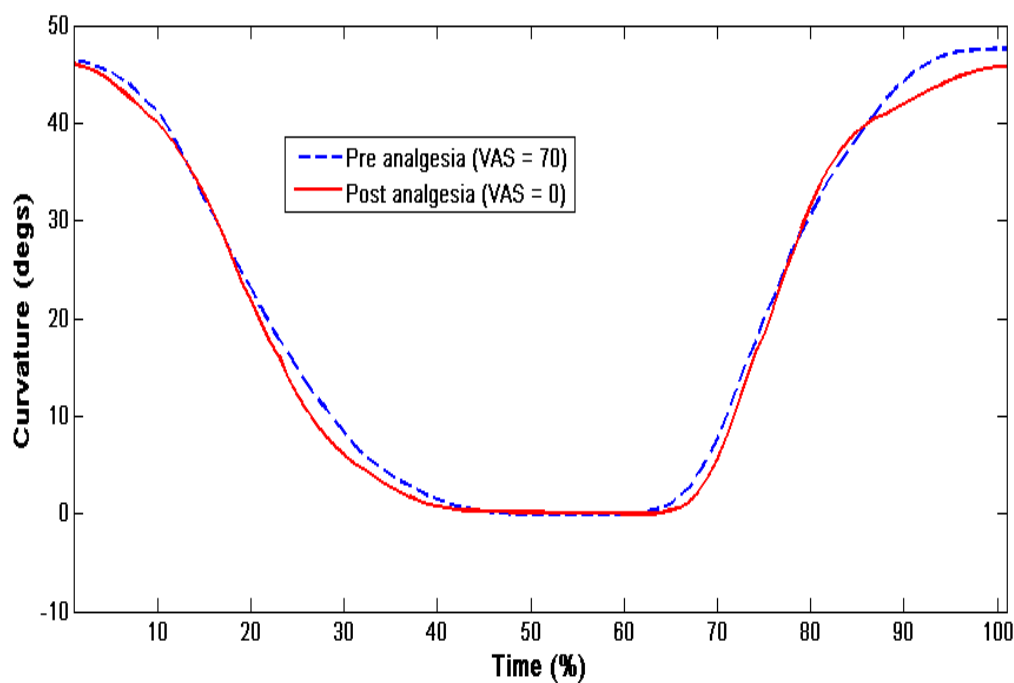
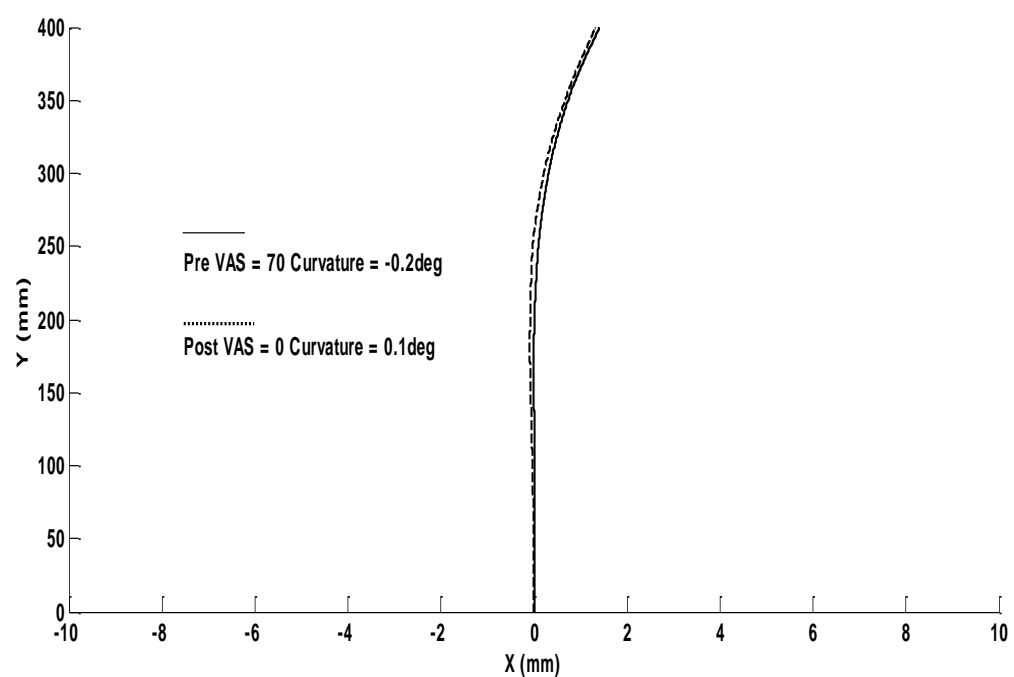


Fig.3.3.1. Medication choices by participants. Combination, 3x Co-codamol and Ibuprofen, 2x Co-codamol and Diclofenac, 1x Paracetamol and Ibuprofen; CLBP Combination, Codydramol and Naproxen

A single participant's curvature, for the movement of flexion, before and after pain relief is presented in figure 3.3.2 (a) along with the same individuals sagittal profile at the end of range of the movement (figure 3.3.2 (b)).



(a)



(b)

Figure 3.3.2 (a) and (b). An individual's curvature throughout the movement of flexion (a). The same individual's sagittal profile at end range flexion.

The group data demonstrates that all ALBP participants, except one, reduced their peak kyphotic curvature following pain relief for flexion and lifting. A significant reduction in peak curvature was demonstrated following pain relief for flexion and lifting in the whole lumbar spine, and for lifting in the lower lumbar spine, in the ALBP group (table 3.3.4). This reduction demonstrates less kyphosis at the peak of flexion and lifting.

No significant difference was observed for extension in the ALBP group and the CLBP demonstrated no significant difference for any movement or region of the spine (table 3.3.4).

The ALBP group demonstrated that 12.5% of individuals were defined as responders to pain relief along with 15.6% for the lower lumbar peak curvature. The CLBP group demonstrated 20% of individuals increased their whole lumbar curvature following pain relief along with 16.7% for the lower lumbar curvature. Chi squared testing demonstrated that neither group were more likely to increase their peak curvature following pain relief ( $\chi^2 = 0.64$ ,  $p=0.422$  for whole and  $\chi^2 = 0.01$ ,  $p = 0.911$  for the lower lumbar spine).

Sequencing demonstrated that in the ALBP group 27%, 17% and 22% of individuals altered the quartile of greater change for flexion, lifting and extension respectively. In the CLBP 50%, 22% and 62% of individuals shifted the quartile of greatest change following pain relief for flexion, lifting and extension respectively. Chi squared testing demonstrated that neither group were more likely to alter their curvature sequencing overall ( $\chi^2 = 1.649$ ,  $p=0.199$  for whole and  $\chi^2 = 0.343$ ,  $p = 0.558$  for the lower lumbar spine).

Table 3.3.4 Mean (sd) peak curvatures before and after analgesia (degrees).

ALBP	Whole lumbar spine			Lower lumbar spine		
	Pre	Post	Diff	Pre	Post	Diff
Flexion						
Curvature (°)	<b>-5.4 (9.8)</b>	<b>-2.3 (9.8)</b>	<b>3.1 (2.1)*</b>	-3.3 (7.0)	-2.2 (6.2)	1.1 (2.0)
Lifting						
Curvature (°)	<b>-1.3 (13.5)</b>	<b>1.4 (12.9)</b>	<b>2.7 (2.6)*</b>	<b>-0.8 (8.0)</b>	<b>1.4 (8.5)</b>	<b>2.2 (2.9)*</b>
Extension						
Curvature (°)	40.1 (17.5)	45.3 (16.6)	5.2 (9.9)	26.3 (11.3)	29.7 (12.8)	3.4 (9.7)
CLBP	Whole lumbar spine			Lower lumbar spine		
	Pre	Post	Diff	Pre	Post	Diff
Flexion						
Curvature (°)	-13.1 (8.1)	-12.7 (7.7)	0.4 (4.8)	-6.4 (4.9)	-6.7 (5.7)	-0.3 (3.4)
Lifting						
Curvature (°)	-13.5 (9.4)	-11.8 (11.9)	1.6 (3.6)	-5.7 (6.5)	-5.2 (9.0)	0.5 (3.6)
Extension						
Curvature (°)	46.5 (11.9)	45.7 (12.2)	-0.8 (6.0)	30.0 (14.7)	29.3 (14.4)	-0.6 (5.5)

\*  $p < 0.05$ . Pre, pre-pain relief; Post, post-pain relief; Diff, mean difference between pre and post values.

### 3.3.5 Discussion

The present study examined the effects of pain relief on lumbar curvature and sequencing behaviour in acute and chronic low back pain sufferers. The results show that pain relief did not attenuate the lumbar curvatures for acute or chronic LBP sufferers except for flexion and lifting in the acute LBP group. The significant effects noted for flexion and lifting demonstrated a reduction in the amount of lumbar kyphosis curvature achieved.

It is commonplace to use the difference between standing curvature and end range curvature in the measurement of ROM therefore despite this being the first study to investigate the effects of pain relief on lumbar curvature directly; others have studied its effect on ROM which

may therefore be used to imply curvature and aid comparison. The current findings are consistent with previous studies which showed pain relief did not increase flexion or extension ROM (Lilius et al., 1989). These authors demonstrated significant pain-relief, induced by facet joint injection, in the order of 18.3 mm on VAS, did not result in gains in flexion or extension ROM. The findings of the current study are in contrast to Jarzem et al., (2005), who did report a significant gain in ROM following pain-relief. Jarzem et al., (2005) demonstrated that following the application of TENS, LBP sufferers had a significant reduction in pain and an increase in ROM during flexion and extension as measured by an inclinometer. The authors presented only p-values therefore the actual magnitude of gains cannot be determined, however the authors allude to the magnitude being small and within the realms of measurement error (Jarzem et al., 2005). Furthermore the measurement technique failed to take into account the pelvis contribution therefore it is possible that small changes in ROM could be due to changes in the pelvis motion.

It is well documented that LBP sufferers often display reduced range of motion (Shum et al., 2005a; 2005b; 2007a) which has been suggested as a compensation response to avoid the provocation of pain, however this has been challenged by the results of the present study. If the reduction in curvature (which would result in a reduction in ROM) observed in LBP was a protective response then the partial removal of pain, as achieved in this study, would be expected to result in an increase in curvature. This was not found to be the case suggesting either insufficient pain relief was achieved or that the reduction in curvature seen in sufferers of LBP may be caused by an alternate mechanism.

It should however be remembered that the aim of the present study was to investigate the immediate effects of pain relief. It could be argued that as the movement becomes less painful individuals gradually alter their curvature over time. This may represent some learned response to the reduction of evoked pain similar to that observed in an experimental pain



study (Moseley et al., 2005). The three repeated trials showed no evidence for this gradual change, however it is interesting that the ALBP did reduce their curvature during the less painful testing. This could be explained by a learned response where individuals were predicting further movement evoked pain causing 'movement adaptation' within the experimental environment to achieve less curvature. However the converse could also be true where the reduced curvature in the acute group resulted in less pain and that the administration of analgesia had little overall effect. If this was the case then the movement of extension would not have demonstrated less pain following analgesia. It could also be argued that the overall magnitude of change in peak curvature was small and clinically insignificant. No guidelines exist as to suggest what a clinically significant amount of curvature change might be, however the mean absolute difference observed within an individual over three trials using this measurement method has been reported as <2 degrees (Williams, Haq, & Lee, 2012) suggesting a curvature change of 3 degrees, as seen in this study, is not much greater than natural movement variability.

Lumbar sequencing behaviour has not been extensively studied in the literature therefore the current findings are somewhat novel. The current study demonstrated that around a quarter of ALBP sufferers altered their pattern of lumbar sequencing during flexion and extension in response to pain relief, whereas this was observed in around half of CLBP sufferers.

It may be possible that due to the short duration of pain, the ALBP sufferers demonstrated the same sequencing behaviour as to prior to the pain episode. This would result in pain not causing a change in normal habitual sequencing behaviour and therefore pain relief also results in no alteration in sequencing. However the CLBP group demonstrated a trend towards an increased likelihood of change in sequencing (although not significant with Chi-squared testing), suggesting perhaps that having pain for a significant duration results in an alteration in lumbar sequencing. This could be due to the CNS 'experimenting' to find a more comfortable movement sequence or be reflective of continued adaptation to nociceptive input

over time. Previous authors have suggested that lumbar sequencing is an important variable to assess within the clinical environment for CLBP (O'Sullivan, 2000). Case-study data has shown a reduction in pain in response to change in lumbar sequencing. Unfortunately this was only part of the 'intervention' and therefore it is unclear whether sequencing change in this case was in response to pain relief or was the actual driver of pain relief (Dankaerts et al., 2007). Therefore the cause and effect relationship regarding pain and lumbar sequencing in CLBP remains unclear.

Limitations to this study include the degree of pain relief achieved by the analgesia was less than anticipated. It was assumed that painkillers would relieve all pain associated with movement, however this was not the case. It is not clear if full pain relief in all participants would affect the results. The sample contains individuals with pain confined to the lower back only and therefore the results cannot be extrapolated to other presentations such as those with neurological signs and symptoms. The analysis here is confined to movements in the sagittal plane and it is not known how pain relief affects movements other than flexion, extension and lifting.

### **3.3.6 Conclusion**

This study demonstrates that partial pain relief resulted in no gains in lumbar curvature providing contradiction to the commonly held belief that curvature is modified to minimise the provocation of pain. These results suggest that treatment methods achieving partial pain relief will not result in an immediate gain in lumbar curvature. Pain-relief appears to be no more likely to alter peak curvature or sequencing behaviour in ALBP than CLBP sufferers.

### 3.4 Summary of results and key findings

The results of this chapter demonstrate that the fibre-optic ribbon offers a viable option for the dynamic measurement of sagittal spinal curvature. The repeated measures reliability has been demonstrated in asymptomatic individuals where comparison of the curvature-time curves illustrates excellent consistency of the device despite the natural biological variation associated with repeated movements. The RMSE value provides an estimation of variability associated with both biological variation and measurement error across the whole movement. The RMSE values were small at  $<2.1^{\circ}$  suggesting this measurement system would be reliable enough to detect small changes in curvature and warrants testing in clinical populations.

In clinical populations the reliability results demonstrate excellent consistency in curvature measurement across time for both acute and chronic LBP groups. The RMSE values were also small,  $<3.4^{\circ}$  for lifting in the CLBP group, with the rest of the movements displaying  $<2.8^{\circ}$  RMSE. The reliability of peak curvature measurement demonstrates that this method is reliable for the quantification of this variable also in LBP sufferers with mean absolute errors associated with repeated movements  $<2.1^{\circ}$  across all movements tested (flexion, extension and lifting).

The fibre-optic system has been shown also to be valid by demonstrating excellent similarity to an accepted gold standard in the form of a video-based optoelectronic system. The comparison of curvature-time curves between the two systems illustrated excellent curve similarity and RMSE values of  $<3.5^{\circ}$  even though the data collections were not simultaneous. The values presented are similar to other systems discussed in table 2.1, however this system demonstrates benefits in its ability to be used in any environment and can overcome the limitation of static measurement associated with previous curvature measurement systems. It is however limited to the sagittal plane currently which is due primarily to the current product design. Furthermore as the fibre-optic system captures the entire length of the spine it is possible to calculate curvature for any region of the spine in a single data capture. These

studies demonstrate this function by providing data for the whole lumbar spine (S1-L1) and lower lumbar spine (S1-L3) however it is likely that any region covered by the system could be measured.

The results within this chapter demonstrate that ALBP sufferers achieve significantly less lumbar kyphotic curvature during flexion and lifting for the whole lumbar spine and lifting only for the lower lumbar spine. Sequencing of curvature change has not been extensively studied in relation to LBP. The results demonstrate that this variable can be investigated with this motion analysis device. The quartile of greatest change was most often the second during flexion and lifting and the first quartile for extension in both groups. Matching between quartiles of greatest change for the whole and lower lumbar spine occurred most often during flexion for both groups and extension for the CLBP group.

The results of this chapter demonstrate that partial pain relief does not attenuate lumbar curvature in CLBP sufferers. The pain relief achieved did cause a reduction in peak kyphosis curvature for ALBP sufferers during flexion and lifting. Analysis of responders showed that ALBP sufferers were no more likely to respond to pain relief than CLBP sufferers.

The results following pain-relief suggest around a quarter of participants are likely to alter their quartile of greatest change in the ALBP whereas around a half are likely to shift the quartile of greatest change for flexion and extension in the CLBP group.

#### Summary of key findings

- The fibre-optic system is a reliable and valid method for dynamic curvature measurement.
- ALBP sufferers achieve less peak curvature during flexion and lifting compared to CLBP sufferers for the whole lumbar spine.
- ALBP sufferers achieve less peak curvature during lifting compared to CLBP sufferers for the lower lumbar spine.
- Sequencing of curvature change can be studied using the fibre-optic system.

- The second quartile demonstrated the greatest curvature change during flexion and lifting and the first quartile for extension for both ALBP and CLBP groups.
- Partial pain relief results in a reduction in peak curvature for the ALBP group during flexion and lifting.
- Partial pain relief does not alter any peak curvatures in the CLBP group.
- ALBP sufferers were no more likely to respond to pain relief than CLBP sufferers.
- Around a quarter of ALBP participants alter their quartile of greatest change following pain relief.
- Around a half of CLBP participants alter their quartile of greatest change during flexion and extension and a quarter for lifting.

# **Chapter 4:**

## **Lumbar kinematics**

## **4.1 Higher order kinematic measurement in LBP. Reliability and spatial domain analysis**

### **4.1.1 Introduction**

Previous chapters have described the feasibility of using a novel fibre-optic device for the measurement of lumbar curvature. The limitations have been clearly outlined previously but, briefly include an inability to resolve out of sagittal plane movements. Therefore it is not an ideal solution for the clinician who is trying to measure spinal motions in other planes. There are many options available however most clinical systems are not able to measure dynamic movement in three dimensions (Battie et al., 1987; Bierma-Zeinstra et al., 2001; Dolan et al., 1995; Mannion et al., 2004). Laboratory systems which are able to overcome this limitation, are often expensive and have environmental constraints therefore other options are required to meet the needs of the clinician (Milne et al., 1996; Ng et al., 2009). Furthermore lumbar kinematic reporting has traditionally been limited to ROM, however deficits in higher order kinematics of angular velocity and angular acceleration are more pronounced in LBP sufferers (Marras et al., 1999; Marras et al., 1995; Marras et al., 1986; Novy et al., 1999; Shum et al., 2005a; 2005b; , 2007a). A clinical motion analysis system therefore should not have environmental constraints, be able to resolve dynamic three dimensional movements and be able to provide reliable measurement of higher order or differential kinematics.

The resolution of angular velocity measurement would enable the analysis to extend to the spatial domain. Spatial plots have been used previously to present the relationship between ROM and angular velocity (Burgess-Limerick, Abernethy, & Neal, 1993; Lee & Wong, 2002; Li, van den Bogert, Caldwell, van Emmerik, & Hamill, 1999b). Such a method could be used to visualise the degree of movement irregularity associated with the movement trajectory, however to date no method has been described which quantifies the movement irregularity as a variable in LBP.

The above approach could be achieved using inertial sensors, however before such technologies can be recommended for routine clinical use, reliability should be ascertained pertaining especially to the differential kinematics and spatial domain analysis.

#### **4.1.2 Aim of study**

The aim of this study was to investigate the repeated measures reliability of an inertial sensor system to measure differential kinematics of LBP sufferers in a clinical environment and to describe the use of spatial plots, including the quantification of motion irregularity, to study the spatial relationship between variables.

#### **4.1.3 Methods**

##### **4.1.3.1 Participants**

Forty LBP sufferers were initially recruited (twenty acute and twenty chronic) from general practitioner referrals to local therapy departments. Referrals were screened and appropriate individuals assessed for inclusion and exclusion criteria. Inclusion criteria included LBP confined to between the lower ribs and inferior gluteal folds, pain evoked by at least three of the test movements and aged between 18 and 55 years old. Exclusion criteria included leg symptoms (pain, paraesthesia and anaesthesia), history of spinal fracture, tumour or surgery. Four ALBP and eight CLBP sufferers were excluded as they did not demonstrate three or more movements which evoked pain. Mean pain scores in the week preceding data collection were measured using visual analogue scale (VAS) and participants also completed a Tampa scale of kinesiophobia questionnaire. Participant demographics are presented in table 4.1.1.

This study was approved by the National Research Ethics Service of the National Health Service and written informed consent was obtained following a standardised explanation of procedures and risks.



Table 4.1.1. Participant demographics (mean (sd)).

	ALBP	CLBP
Male/Female	10/6	7/5
Age (years)	42.6 (7.2)	34.5 (10.0)
Height (m)	173.5 (10.1)	174.2 (11.7)
Weight (kg)	83.7 (15.3)	82.0 (13.2)
BMI (kg/m <sup>2</sup> )	27.7 (3.5)	27.4 (4.8)
TSK	39.3 (4.1)	38.3 (7.5)
VAS (mm)	61.5 (18.9)	45.7 (22.1)
Duration (days)	12.0 (7.3)	4030.2 (2992.2)

BMI, Body Mass Index; TSK, Tampa Scale of Kinesiophobia; VAS, Visual Analogue Scale.

#### 4.1.3.2 Instrumentation

Two wired 3DM-GX3-25 inertial sensors were used to measure lumbar kinematics (Microstrain, VT, USA). Each sensor contained three orthogonally aligned integrated sensing elements including gyroscopes ( $\pm 300^\circ/\text{s}$ ), accelerometers ( $\pm 5\text{g}$ ) and magnetometers. Each sensor consisted of a durable casing with dimensions 44mm (h) x 25mm (w) x 11mm (d), and weight of 18g. The sensors have a reported accuracy of  $\pm 0.5^\circ$  during static testing and  $\pm 2.0^\circ$  during dynamic (i.e. cyclic) testing.

One sensor was placed over the S1 spinous process and the second over the L1 spinous process. The sensors were attached using double sided tape with the wires secured to the trunk so as not to move the sensors erroneously. The sensors were connected to a purpose built datalogger and software (ThetaMetrix, UK) and data were collected at 100Hz. The relative orientations between the L1 and S1 sensor were determined from the direction cosine matrices (Burnett et al., 1998; Grood & Suntay, 1983; Lee et al., 2003). Flexion, left side bending and left rotation were considered positive and the opposite movements negative.

#### 4.1.3.3 Procedure

Participants stood upright with feet positioned naturally along a line before completing three trials of flexion, extension, left side bending, right side bending, left rotation, right rotation and a box lift. The box was positioned on markers ensuring identical placement and measured 460 x 260 x 300mm and weighed 3kg. The movement order was identical for all participants and the sensors remained attached throughout.

#### 4.1.3.4 Data analysis

All raw data were transferred to Matlab for processing (Mathworks, R2008b). The movement-time data for each movement were determined and smoothed using a fourth order zero-lag Butterworth filter with a cut off frequency of 1Hz (Tsang et al., 2011). The movement-time data were differentiated to yield the angular velocity and double differentiated to calculate angular acceleration. The curves for each individual movement were time-normalised with respect to the x-axis (time) so that each movement had the same time base from 0% to 100%. This normalisation process takes into account the differing rates at which the participants completed the trials and allows the direct comparison of the kinematic pattern. The coefficient of multiple correlation (CMC) (Li & Caldwell, 1999; Williams et al., 2010a) and root mean square error (RMSE) were calculated from the normalised displacement, angular velocity and angular acceleration curves of the primary movement, to determine the similarities between the three movement trials. The peak range of motion, peak angular velocity and peak angular acceleration values were obtained and intra-class coefficients ( $ICC_{3,1}$ ) calculated along with the mean absolute differences between peak values for the three trials. Two peak values were obtained, positive and negative. Positive angular velocity and angular acceleration relate to flexion, left side bending and left rotation movement, whereas negative angular velocity and angular acceleration relate to movement in the opposite direction. Interpretation of relative reliabilities was based on the criterion by Portney and Watkins (2000) where  $>0.75$  suggests good reliability,  $>0.5$  suggests moderate reliability and  $<0.5$  suggests poor reliability, however

error percentages were also calculated to aid in the clinical interpretation. Angular velocity-angular displacement plots were used to reveal the spatial relationship where the overall shape of the plot describes the trajectory of the movement and dynamic control. The movement irregularity was determined from the plots by separating the plot into quartiles and fitting the data with a 4<sup>th</sup> order polynomial. The quartiles were defined as the region from motion onset to peak angular velocity (quartile 1); peak angular velocity to end of range of motion or maximum angular displacement (quartile 2); maximum angular displacement to minimum angular velocity (quartile 3) and minimum angular velocity to minimum angular displacement or return to upright standing (quartile 4). The beginning and end of the polynomial was fixed to match the collected data and the polynomial was used to determine the RMS difference between the data and the polynomial and therefore provide quantification of motion irregularities of each quartile. The RMS was normalised to the peak angular velocity of each individual to provide a value representing the percentage of peak angular velocity. This was necessary to avoid those with small overall velocities demonstrating comparatively small RMS values despite potentially less smooth motions. Scores of motion irregularity (RMS) were determined for each quartile across each movement and  $ICC_{(1,1)}$  and mean absolute difference between trials were used to determine reliability of such a method with mean scores used to compare ALBP and CLBP sufferers.

#### **4.1.4 Results**

Mean (sd) for displacement, angular velocity and angular acceleration for each movement can be found in table 4.1.2 and a typical repeated trials kinematics graph is presented in figure 4.1.1.

The mean CMC values were found to be good across all movement-time and angular velocity-time curves (table 4.1.3) which demonstrates highly similar movement patterns across

repeated movements. The mean CMC values for angular acceleration-time curves show moderate to good consistency across the different movements tested (table 4.1.3). The RMSE magnitudes across all kinematic variables with respect to time were small for all repeated movements (figure 4.1.2).

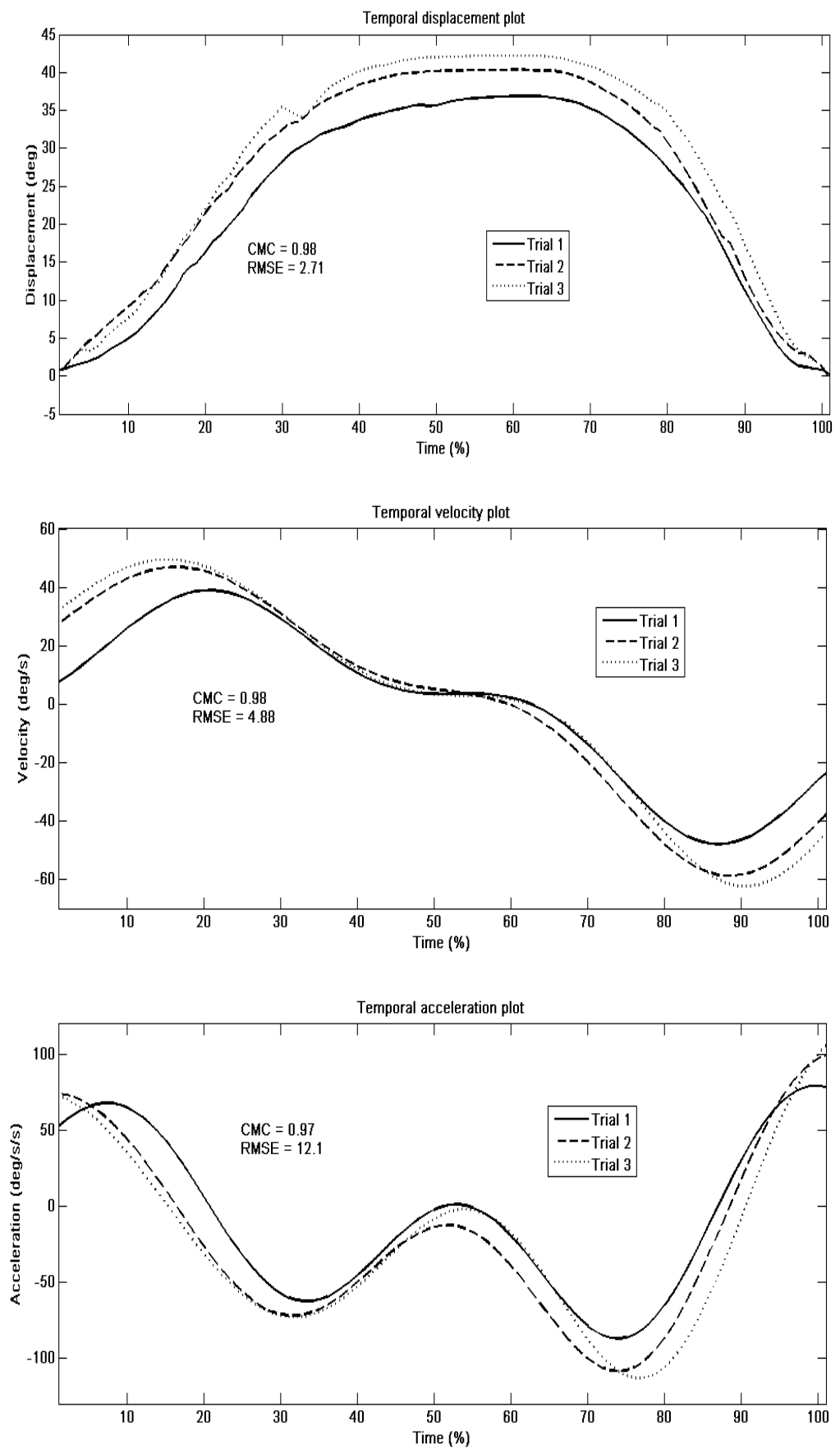


Figure 4.1.1. A typical kinematics graph of repeated trials for a single participant.

Table 4.1.2. Mean (sd) values of kinematic variables recorded by the inertial system

	Disp ( $^{\circ}$ )		+ve Vel ( $^{\circ}\text{s}^{-1}$ )		-ve Vel ( $^{\circ}\text{s}^{-1}$ )		+ve Acc ( $^{\circ}\text{s}^{-2}$ )		-ve Acc ( $^{\circ}\text{s}^{-2}$ )	
	ALBP	CLBP	ALBP	CLBP	ALBP	CLBP	ALBP	CLBP	ALBP	CLBP
Flex	35.1	48.1	20.1	28.2	20.1	29.0	29.8	34.1	26.6	41.3
	(14.5)	(11.4)	(15.8)	(9.6)	(11.8)	(10.0)	(18.8)	(28.7)	(16.7)	(21.0)
Ext	11.8	16.5	10.6	15.0	6.2	10.0	18.9	25.1	14.6	26.1
	(8.8)	(5.1)	(6.3)	(6.3)	(3.0)	(4.3)	(10.1)	(12.0)	(8.1)	(14.5)
LSF	11.5	14.6	8.2	9.6	11.2	11.6	15.3	18.9	17.5	17.6
	(3.7)	(6.9)	(4.4)	(5.8)	(4.3)	(5.8)	(8.5)	(11.4)	(7.6)	(10.8)
RSF	15.0	15.9	13.4	13.6	8.8	10.2	20.9	20.6	17.2	20.7
	(4.1)	(4.6)	(4.8)	(5.9)	(3.7)	(5.1)	(9.3)	(11.8)	(8.2)	(11.3)
LRot	7.1	4.4	5.0	4.5	5.1	5.0	11.5	13.5	10.6	13.5
	(3.5)	(3.6)	(2.6)	(3.1)	(2.6)	(3.9)	(4.5)	(9.2)	(4.6)	(8.9)
RRot	10.6	8.3	7.0	6.1	5.6	6.4	12.2	14.9	12.7	14.2
	(4.5)	(4.1)	(3.0)	(4.2)	(2.5)	(4.8)	(7.0)	(10.1)	(5.7)	(9.9)
Lift	29.9	43.8	21.5	33.8	26.6	36.3	44.1	59.7	42.8	58.5
	(15.0)	(12.6)	(14.0)	(13.9)	(13.5)	(14.5)	(23.3)	(29.8)	(24.6)	(30.5)

Flex, Flexion; Ext, Extension; LSF, Left side flexion; RSF, Right side flexion; LRot, Left rotation; RRot, Right rotation; Disp, Angular displacement; +ve, positive; -ve, negative; Vel, Angular velocity; Acc, Angular acceleration.

Table 4.1.3. Mean (sd) CMC and ICC values across each movement

	CMC			ICC				
	Disp	Vel	Acc	Disp	+ve Vel	-ve Vel	+ve Acc	-ve Acc
ALBP								
Flex	0.97 (0.02)	0.93 (0.04)	0.72 (0.14)	0.99	0.93	0.94	0.96	0.95
Ext	0.85 (0.09)	0.83 (0.08)	0.61 (0.18)	0.99	0.94	0.89	0.88	0.88
LSF	0.94 (0.05)	0.93 (0.04)	0.77 (0.10)	0.97	0.93	0.94	0.92	0.88
RSF	0.97 (0.02)	0.94 (0.04)	0.78 (0.14)	0.99	0.93	0.94	0.94	0.92
LRot	0.86 (0.08)	0.84 (0.10)	0.67 (0.13)	0.92	0.93	0.80	0.76	0.71
RRot	0.83 (0.08)	0.89 (0.06)	0.67 (0.16)	0.95	0.93	0.95	0.88	0.89
Lift	0.92 (0.04)	0.90 (0.04)	0.71 (0.11)	0.99	0.97	0.98	0.95	0.96
CLBP								
Flex	0.94 (0.04)	0.90 (0.06)	0.66 (0.13)	0.97	0.92	0.93	0.93	0.89
Ext	0.91 (0.05)	0.89 (0.05)	0.65 (0.18)	0.94	0.93	0.93	0.96	0.83
LSF	0.94 (0.03)	0.91 (0.04)	0.73 (0.10)	0.99	0.91	0.98	0.93	0.88
RSF	0.94 (0.03)	0.91 (0.07)	0.70 (0.26)	0.97	0.98	0.94	0.94	0.94
LRot	0.72 (0.23)	0.71 (0.18)	0.52 (0.25)	0.9	0.96	0.96	0.90	0.91
RRot	0.75 (0.26)	0.81 (0.15)	0.59 (0.24)	0.97	0.97	0.96	0.95	0.95
Lift	0.92 (0.05)	0.88 (0.06)	0.67 (0.11)	0.99	0.96	0.95	0.90	0.88

Flex, Flexion; Ext, Extension; LSF, Left Side-Flexion; RSF, Right Side-Flexion; LRot, Left Rotation; RRot, Right Rotation; +ve, positive; -ve, negative; Vel, Angular velocity; Acc, Angular acceleration.

Mean ICC values were shown to be good across all movements for both groups (table 4.1.3)

and mean absolute differences of repeated peak measures were small, below  $3.7^{\circ}$ ,  $5.6^{\circ}\text{s}^{-1}$  and  $13.5^{\circ}\text{s}^{-2}$  for angular-displacement, angular velocity and angular acceleration respectively (figure 4.1.2).

In order to assist in the interpretation of these errors, percentage absolute errors were calculated demonstrating mean errors of 11%, 23% and 28% for ALBP and 15%, 20% and 28% for CLBP for angular displacement, angular velocity and angular acceleration respectively.

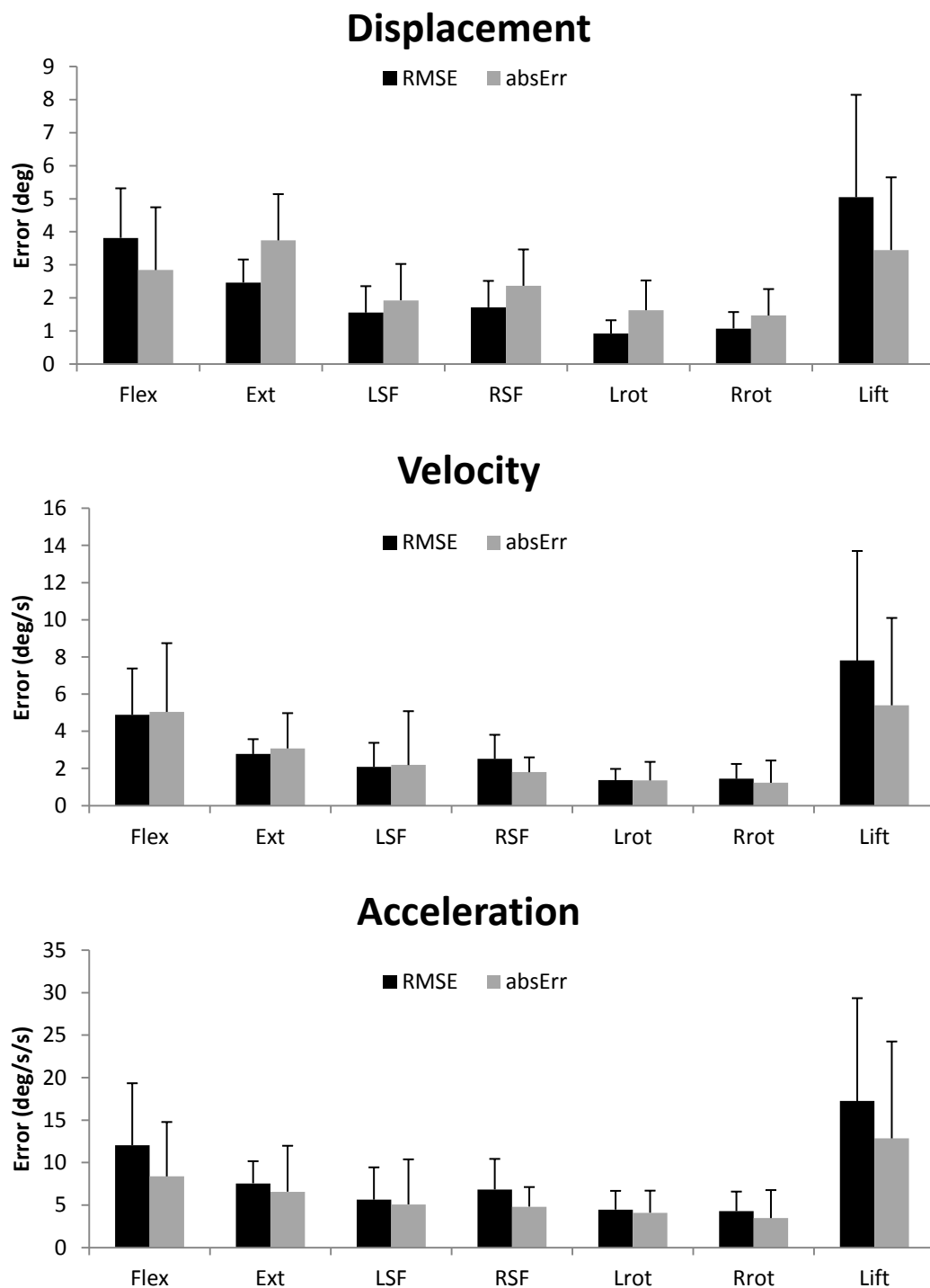


Figure 4.1.2. Root Mean Square Error values for Angular displacement- Angular velocity- and Angular acceleration-time curves along with absolute mean difference in peak values of Angular displacement, Angular velocity and Angular acceleration (Flex, Flexion; Ext, Extension; LSF, Left Side-Flexion; RSF, Right Side-Flexion; Lrot, Left Rotation; Rrot, Right Rotation).



The spatial plots demonstrate similarity in the trajectory of the movement as all movements show consistency in the graphical shape produced. Figure 4.1.3 enables the visualisation of the spatial relationship of angular displacement and angular velocity during forward bending. The movement begins on the left near the origin. The forward bending commences and proceeds in the upper half of the graph until a maximal angular displacement of around  $50^{\circ}$  was reached. This was achieved with a peak angular velocity of around  $30^{\circ}/s^{-1}$  evident just before half of the total angular displacement was reached. Full flexion (far right of the graph) is easily detectable as the point of maximal angular displacement and point of zero angular velocity. Return from flexion occupies the lower half of the graph. It is clear that for any point in time the motion of the spine can be defined relative to the angular displacement or angular velocity.

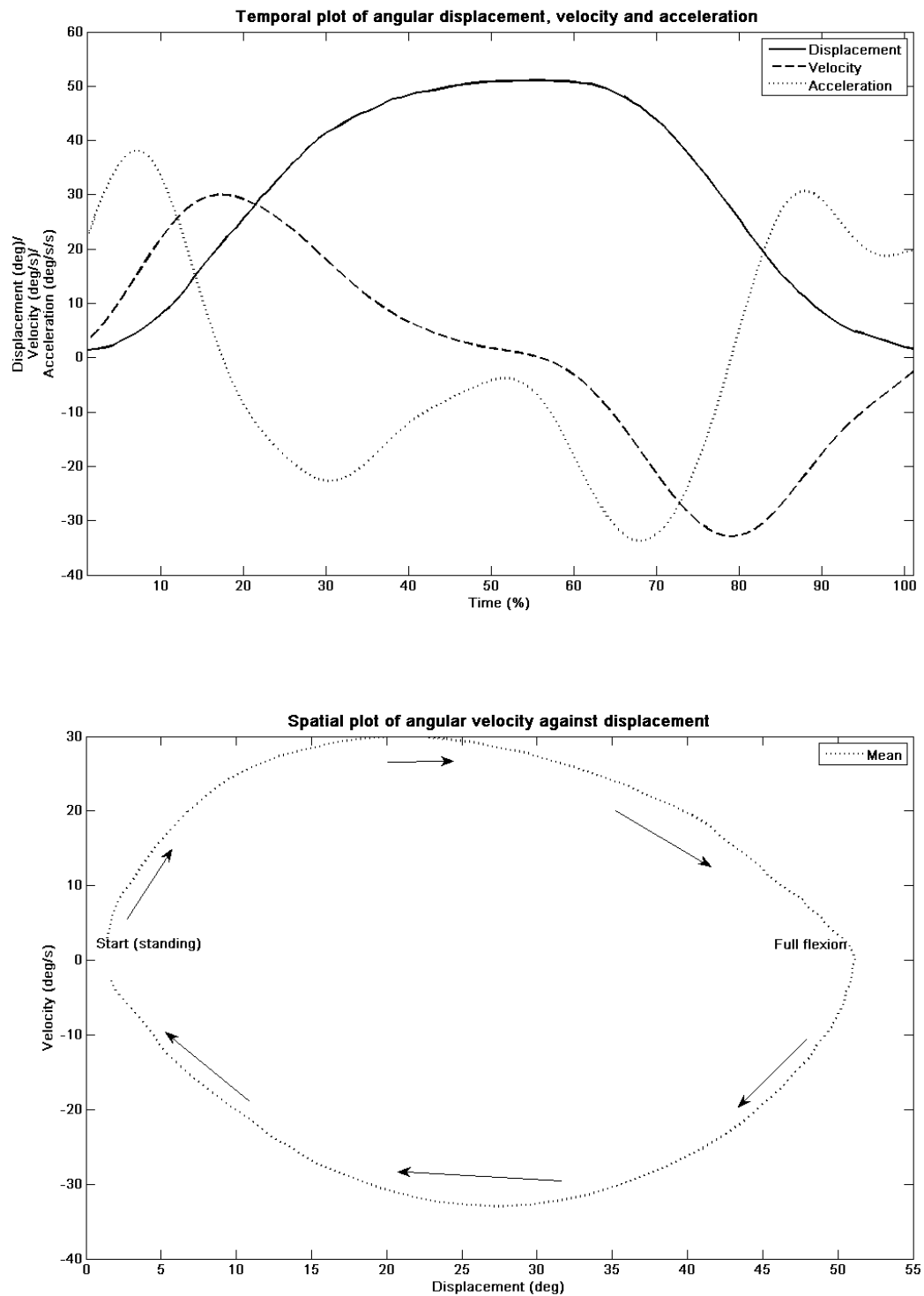


Figure 4.1.3. (a) Plot displays temporal relationship of angular displacement, angular velocity and angular acceleration against time. Plot (b) visually displays the relationship between the angular displacement and angular velocity.

A typical graphical representation of the motion irregularity calculation is presented in figure 4.1.4. The graphs demonstrate the curve fitting process and resultant RMS calculation to determine a score of irregularity. The similarities in irregularities between repeated trials was moderate to good for all the movements investigated and mean absolute differences small (table 4.1.4) suggesting good reliability of such a method.

Table 4.1.4. ICC and mean absolute differences between repeated trials

	ALBP		CLBP	
	ICC	absDiff	ICC	absDiff
Flexion	0.73	1.8 (1.9)	0.50	1.4 (1.1)
Extension	0.49	5.4 (7.1)	0.51	2.5 (2.4)
LSF	0.79	1.9 (2.5)	0.68	2.0 (2.2)
RSF	0.86	1.2 (1.6)	0.71	1.8 (2.4)
LRot	0.65	3.7 (3.9)	0.75	5.5 (5.4)
RRot	0.61	3.2 (2.9)	0.83	4.3 (5.0)

LSF, left side flexion; RSF, right side flexion; LRot, left rotation; RRot, right rotation; absDiff, mean absolute difference.

It is evident from graph 4.1.4 (a) that the first and fourth quartiles demonstrate small movement irregularity with low values whereas the second quartile displays greater irregularity especially at more than 40 degrees angular displacement. It is also around this region that more irregularity is evident for quartile 3 resulting in a score of 2.6. If this graph is compared to (b) then it is evident that quartiles 1 and 4 are again those with fewest irregularities, as mirrored by the group results (table 4.1.5) however the second demonstrates greater irregularity compared to the other quartiles (score = 8.5). This is in part due to some jerkiness visualised in the final 5 degrees of angular displacement. Quartile 3 also scores highly due to motion irregularity in the first 5 degrees of returning from flexion.

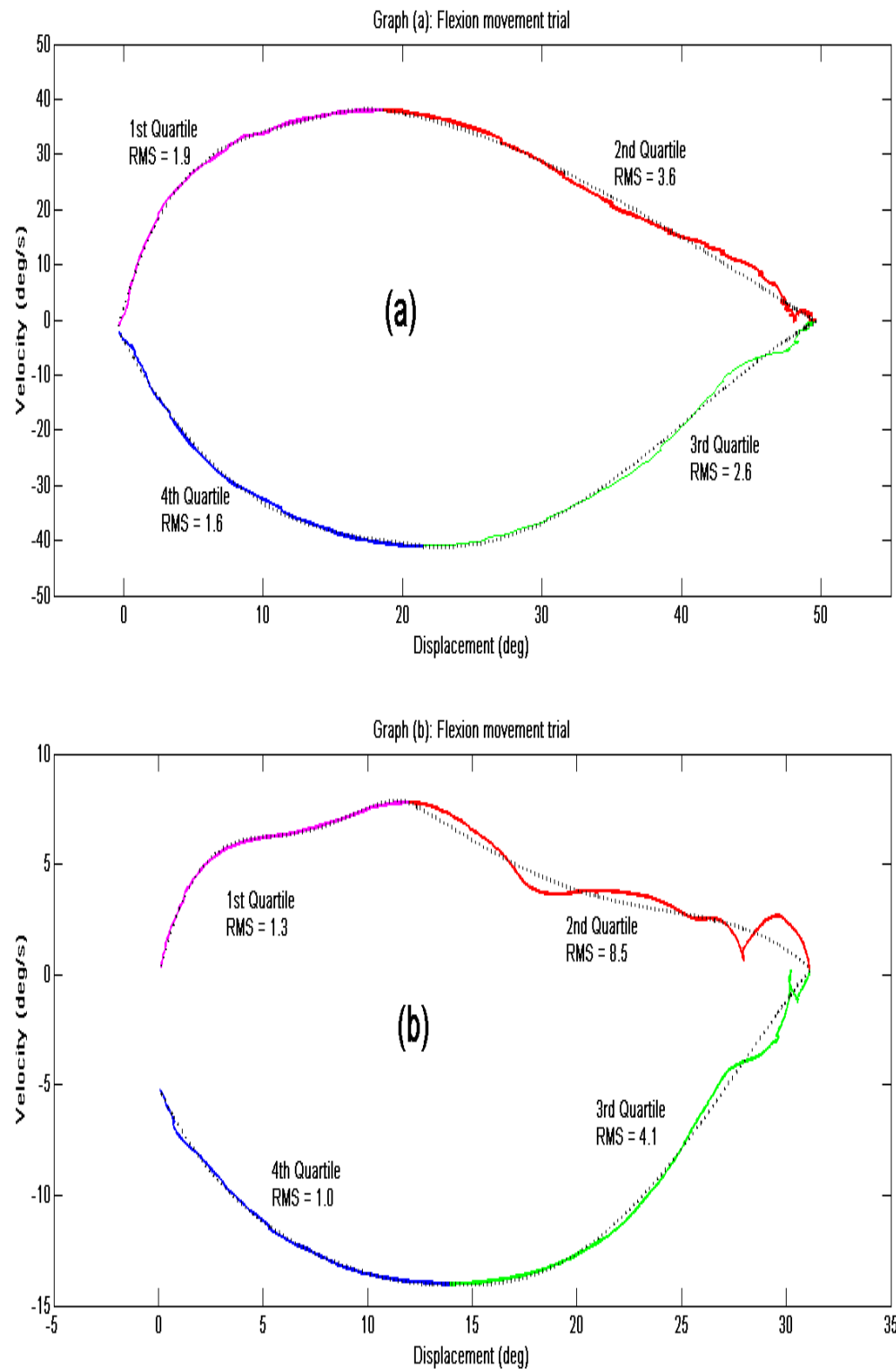


Figure 4.1.4. Quantification of spatial plots. (a) Flexion trial of individual participant. (b) Flexion trial of different individual. Quartiles are coloured and polynomials are presented with black dotted lines.

The mean numerical motion irregularity score for each quartile for each group as a whole is presented in table 4.1.5 and shows that significant differences in motion irregularity values

were evident in the movements of extension, left and right side flexion, left rotation and lifting (table 4.1.5). The quartile with the greatest irregularity was consistently the second for all movements except left rotation and lifting in the CLBP group. The quartile with the lowest irregularity was consistently the last in all except left and right rotation for the ALBP group and right rotation and lifting for the CLBP group where least irregularity was in the first quartile.

Table 4.1.5. Quantification of movement irregularity per quartile.

	ALBP	CLBP
Flexion		
1	3.1	3.7
2	6.4	4.9
3	4.4	3.7
4	3.5	2.4
Extension		
1	6.0	6.1
2	13.1	6.3*
3	7.2	5.5
4	3.1	4.0
LSF		
1	2.0	5.6*
2	8.0	5.8
3	5.5	3.6
4	1.3	2.6
RSF		
1	1.4	3.5*
2	7.1	7.2
3	4.1	4.2
4	1.1	2.9*
LRot		
1	6.8	16.7*
2	11.4	14.3
3	9.3	13.3
4	11.4	11.5
RRot		
1	5.1	6.4
2	10.5	14.3
3	7.4	11.5
4	5.6	8.8
Lift		
1	3.4	2.9
2	4.7	4.3
3	4.1	3.8
4	2.2	4.5*

\* P = 0.05 or less; LSF, left side flexion; RSF; right side flexion; LRot, left rotation; RRot, right rotation.

#### 4.1.5 Discussion

The results of this study show that inertial sensors are capable of reliably measuring both the temporal and spatial characteristics of spinal kinematics in LBP sufferers, including angular velocity and angular acceleration. The similarities of the movement-time curves and angular velocity-time curves were good for all movements with the exception of rotation. The similarities of repeated rotation movement-time curves and angular velocity-time curves were less than the other movements, however the RMSE of the rotation curves were all less than  $1.3^{\circ}$  and  $0.9^{\circ}\text{s}^{-1}$  respectively. The slighter lower CMC values for rotation suggest the sensors are less consistent in measuring this motion. This may be due to the changing contours of the paraspinal muscles during rotation which could result in small twist errors of the L1 sensor or alterations in the amount of skin movement or slippage between the sensor and skin. However the magnitude of this lumbar movement and its angular velocity were small and therefore small errors may be magnified. In this case it may be more appropriate to use the peak values, which were shown to be more consistent, with an ICC of greater than 0.90 and 0.92 for angular displacement and angular velocity respectively. Furthermore the mean differences between repeated peak measures of rotation range of motion were less than  $1.8^{\circ}$  for angular displacement and  $1.7^{\circ}\text{s}^{-1}$  for angular velocity showing excellent consistency. Peak angular velocity values were highly consistent and little difference between positive (flexion phase, left side flexion phase and left rotation phase) angular velocity and negative angular velocity was observed. This suggests that movements from standing to end range were achieved as consistently as the movements from end of range back to standing, with the exception of extension. The angular velocity from standing to end range extension (negative angular velocity) showed greater variability and it was observed that this movement was completed with less angular velocity compared with the movement from end range extension to standing (positive angular velocity). It is unclear whether this was a method to minimise any evoked pain during extension or whether LBP sufferers have poorer movement awareness into this

direction resulting in more time needed to judge end of range during extension, as has been suggested previously (Williams, Haq, & Lee, 2010b).

The angular acceleration-time curves display good similarity, indicating that this variable is reliable for clinical purposes, however the results are lower than for the other kinematic variables. This may suggest greater angular acceleration movement variation from the participants as biological error, however may also reflect the sensitivity of the double derivative technique to yield angular accelerations. This technique is dependent on the degree of smoothing offered to the original angular displacement data and also any smoothing applied to the angular acceleration data directly. The peak angular accelerations ICC values were good suggesting this variable is very reliable for use in quantifying lumbar angular accelerations in LBP sufferers. Moreover the mean absolute difference in peak value was small, especially for movements other than lifting, typically less than  $10.9^{\circ}\text{s}^{-2}$ .

The results of this study are comparable to those available in the literature (table 4.1.6). The differences that do exist may be due to the different characteristics in the participants, with non-LBP participants displaying greater movement angular velocity and angular acceleration. Previous studies have found excellent reliability for measuring lumbar angular-displacement in non-clinical populations (Lee et al., 2003) with CMC values slightly higher than those achieved in the present study. These differences are may be due to greater intra-participant movement variation possibly due to having or anticipating evoked pain during the movements. This may be especially true for the temporal kinematics where angular acceleration-time curves would be significantly affected by less smooth movements. This may also explain the discrepancy between the CMC values and ICC values found in this study. CMC values provide a measure of how similar the entire curves are across the whole movement, whereas comparisons of single peak values do not take into account the movement pattern across time. Variation in how the participant reaches end of range will significantly affect the CMC values but not the ICC peak values.

Table 4.1.6. Comparison of values from the literature

Study	Methods	Angular velocity ( $^{\circ}\text{s}^{-1}$ )	Angular acceleration ( $^{\circ}\text{s}^{-2}$ )
Marras and Granata (1997)	LMM during lifting Non-LBP group	39-47	79-97
Marras et al., (2000a)	LMM during lifting Non-LBP group	47.9 $\pm$ 14.9	105.9 $\pm$ 9.8
Marras et al., (2001)	LLM during lifting LBP group Non-LBP group	21.3 $\pm$ 18.6 36.5 $\pm$ 28.0	
Marras and Wongsam (1986)*	Potentiometer during flex/ext LBP group Flex phase Return phase Ext phase Return phase Non-LBP group Flex phase Return phase Ext phase Return phase	13-15 11-16 2 6  48 31-33 33 14	
Pal et al., (2007)	Opto-electronic during flex/ext Non-LBP group Flexion phase Return phase	43.5 $\pm$ 7.2 36.7 $\pm$ 8.3	
Milosavljevic et al., (2008)	Opto-electronic during flex/ext Non-LBP group Extension phase Return phase	21.7 $\pm$ 8.6 37.0 $\pm$ 14.7	
Esloa et al., (1996)	Opto-electronic during flex phase History of LBP group No history of LBP group	35.9 $\pm$ 11.1 41.8 $\pm$ 12.6	
McClure et al., (1997)	Opto-electronic during return from flexion phase History of LBP group No history of LBP group	35.2 $\pm$ 10 30.5 $\pm$ 9.5	
Shum et al., (2007a)	Electromagnetic during picking up activity Non-LBP group LBP group LBP + SLR group	30 $\pm$ 11 17 $\pm$ 8.5 19 $\pm$ 8.5	

LMM, Lumbar Motion Monitor; LBP, low back pain; Flex, Flexion; Ext, Extension; SLR, straight leg raise. \* Taken from graph.



The results of this study demonstrate that quantification of motion irregularity using spatial plots is possible and the results, as a score of motion irregularity are reliable. The absolute difference in scores between repeated measurements is usual in determining true difference between groups or conditions. These results demonstrate that the largest differences were evident for the measurement of extension in the ALBP group with similar findings evident for the CLBP group during rotation. This suggests natural intra-individual movement variation is likely for these groups during these motions and serve as a guide in determining true change or difference in motion irregularity between conditions.

The results demonstrate some subtle differences in movement of relative quartiles in ALBP and CLBP sufferers. The ability of the spatial plots to display movement coordination and control is of great use to the clinician. This is the first time such movement profiles have been quantified and applied to LBP sufferers and it is evident that this new information enables the identification of which section of the movement is affected. The quantification of motion irregularity demonstrates a consistent pattern across the LBP sufferers where more irregular motions are evident in the quartile leading up to end of range. This may possibly be due to attempts to minimise provocation of pain. As such it is possible the individual adjusts or 'explores' the movement close to the terminal range in an attempt to find the most comfortable path. This results in deviations in angular velocity behaviour and causes greater scores in the irregularity quantification. However it may also be the case that this loss of movement smoothness represents an impairment of spinal function due to perhaps alteration in proprioceptive input and therefore represents an attempt to increase afferent information to guide the movement pattern. Unfortunately the cause of such discrepancies remains unclear, however the results demonstrate that spatial plots are able to discriminate between movement of greater or less irregularity and that the kinematic variable of motion irregularity maybe of interest in future kinematic studies of LBP sufferers.

#### **4.1.6 Conclusion**

The results of this study show that the inertial sensor system is capable of providing a reliable and novel method of tracking lumbar kinematics in LBP sufferers. Importantly these results show that not just angular displacement but angular velocity and angular acceleration are able to be studied using such sensor technology. The advantages of such a motion analysis system are that it is simple to use, quick to administer and relatively cheap compared to other laboratory based systems. Furthermore the system is fully portable and therefore ideal for use within the clinic, as displayed in this study as all measurements were taken within a clinic environment. The additional kinematic information can be easily incorporated to study the spatial relationship of kinematic variables, which prevent the data collected being limited the temporal characteristics only. Therefore it is possible to study the temporal and spatial kinematics of the lumbar region which may yield important clinical information regarding movement quality and control.

## **4.2 The effect of pain-relief on lumbar kinematics in acute and chronic LBP sufferers**

### **4.2.1 Introduction**

The previous chapter has demonstrated a novel method of clinical analysis of lumbar kinematics extended to differential kinematics and spatial domain analysis. This means that detailed kinematic profiles can be determined in clinical LBP populations within the clinic.

As mentioned previously it has been well documented that LBP sufferers display altered movements compared to individuals without LBP (Marras et al., 2001; Marras et al., 1999; Marras et al., 1986; Shum et al., 2005a; 2005b; Wong et al., 2004). It has been suggested that these are driven by pain (Hodges et al., 2003b), however experimentally induced pain models have shown conflicting results (Arendt-Nielsen et al., 1995; Lamothe et al., 2004; Zedka et al., 1999). Also experimentally induced pain relief studies have demonstrated contradictions in findings, which are probably explainable by variation in task protocol and methods used (Davis et al., 2005; Jarzem et al., 2005; Lilius et al., 1989; Simmonds et al., 2003). It therefore remains unclear as to whether movement alterations in LBP sufferers are driven by pain. Furthermore the effects of pain relief on higher order kinematics and movement irregularities, as quantified through spatial domain analysis, have yet to be reported.

Currently clinicians are faced with difficult decisions regarding optimal treatment for LBP sufferers and management strategies often target pain relief as a way to alter lumbar function, however before such treatments can be rationalised it is imperative to determine the effect of pain relief on lumbar function.

#### **4.2.2 Aim of study**

The purpose of this study was to determine the effects of pain-relief on lumbar kinematics.

This study attempted to address the short comings in the available literature by investigating simple pain-relief as an isolated mechanism, observing its effect on spinal movements in acute and chronic LBP sufferers. It is hoped this information will help clinicians understand the predicted immediate effects of pain targeting interventions.

#### **4.2.3 Methods**

##### **4.2.3.1 Participants**

Forty volunteers were recruited from local physiotherapy and chiropractic clinics. Participants were screened by a physiotherapist for inclusion and exclusion criteria (table 4.2.1) and were divided into one of two groups according to duration of pain (table 4.2.2). There were no significant differences between the groups except for duration and severity of pain. The severity of the pain was rated using a visual analogue scale (0 to 100mm) where participants were asked to mark their average pain over the preceding week, and fear of movement was evaluated using the tampa scale of kinesiophobia questionnaire.

All participants were supplied with information sheets and gave informed consent. The study was approved by the Research Ethics Committee of the National Health Service (08/H1111/38).

Table 4.2.1. Inclusion and exclusion criteria

Inclusion	Exclusion
Pain confined to between lower ribs and inferior gluteal folds	History of tumors
Movement evoked pain	Spinal fractures
Aged 18-55 years old	Surgery
Seeking healthcare for LBP	Neurological signs or symptoms
Acute – Pain present for less than 3 weeks on a history of no pain for at least 12 months	Rheumatological or Neurological disease
Chronic – Pain present on at least 3 days per week for at least 52 weeks	Known spinal deformity

Table 4.2.2. Participant demographics

	ALBP	CLBP	P-value (t-test)
Male/Female	11/9	11/9	
Age (years)	42.7 (6.8)	36.6 (10.8)	0.0754
Height (m)	172.9 (11.3)	173.6 (11.2)	0.6994
Weight (kg)	82.6 (16.6)	83.7 (16.1)	0.5467
BMI (Kg/m <sup>2</sup> )	27.5 (4.0)	26.2 (4.1)	0.4337
TSK	39.0 (4.8)	38.9 (6.9)	0.8541
VAS (mm)	62.2 (16.6)	46 (22)	0.018
Duration	12.3 (6.7) days	9.4 (7.4) years	<0.0001

#### 4.2.3.2 Instrumentation

Two wired 3DM-GX3-25 inertial sensors were used to measure lumbar kinematics (Microstrain, VT, USA). Each sensor contained three integrated sensing elements including gyroscopes ( $\pm 300^\circ/\text{s}$ ), accelerometers ( $\pm 5\text{g}$ ) and magnetometers and provides absolute orientation. One sensor was fixed over the S1 spinous process and the second over the L1 spinous process and were connected to a purpose built datalogger and software (H-Scientific, UK) with data captured at 100Hz. Lumbar spine movement were calculated from the relative orientations between the L1 and S1 sensor. This mathematical technique derives joint angles from direction

cosine matrices and has been described in detail previously (Burnett et al., 1998; Grood & Suntay, 1983; Lee et al., 2003). Flexion, left side bending and left rotation were considered positive and the opposite movements negative.

#### 4.2.3.3 Procedure

Participants were requested to complete three trials of forward and backward bending, side-bending, twisting and lifting. No constraints were placed on any of the movements in order to ensure they were completed naturally. The object to be lifted was a box with dimensions 460x260x300mm which weighed 3kg. No medication was taken on the day prior to testing. The moment of change in their baseline pain was registered by a pain switch pressed by the participant during the movement. The signal generated by the switch marked the data to determine the moment of change in pain. The magnitude of the worst pain evoked by the movements was measured using a visual analogue scale completed following each of the three movement trials.

Participants were then requested to self-administer their usual oral analgesia and given a break of between 45-60 minutes after which the movements were repeated. There were no restrictions on the type of analgesia used and the sensors remained attached throughout the experiment. Only movements which evoked pain were analysed.

#### 4.2.3.4 Data Analysis

All processing was completed using Matlab (Mathworks 2008b). The movement-time data for each movement were determined and filtered using a bidirectional 1Hz low pass filter to remove high frequency noise prior to differentiating and double differentiating to yield angular velocity and angular acceleration respectively (Williams, Haq, & Lee, In Press). The peak range of motion, peak velocities and peak angular accelerations for the primary movements were calculated. Two peak values were obtained for angular velocity and angular accelerations,

positive and negative. Positive angular velocity and angular acceleration relate to when movements are in the direction of flexion, left side bending and left rotation, whereas negative angular velocity and angular acceleration relate to movement in the opposite direction. Using flexion as an example, positive means angular velocity moving towards flexion, negative means a angular velocity moving away from flexion (i.e. extension); and for angular acceleration, positive means an angular acceleration towards the flexion direction and negative means an angular acceleration away from the flexion direction (i.e. extension). To aid interpretation of the change in variables following pain relief, difference values were 'sign' normalised so increases in the variable are positive values and decreases in the variable are negative values. Movement irregularity was also determined by plotting and quantifying the spatial relationships between angular velocity and movement curves. This was achieved using a method described previously (Williams et al., In Press), briefly the angular velocity-movement plot was sectioned into four quartiles and each quartile was fitted with a fourth order polynomial and the normalised root mean square error was used as a measure of movement trajectory irregularity.

#### 4.2.3.5 Statistical Analysis

Statistical analysis was completed using SPSS 20. T-tests were used to compare demographics. A multivariate ANOVA (MANOVA) was completed to determine the effects of group and time on the kinematics, using two independent variables, group (ALBP and CLBP) and time (pre and post pain relief) and five dependent variables (ROM, positive angular velocity and angular acceleration and negative angular velocity and angular acceleration). A MANOVA model was chosen over conducting numerous ANOVAs to reduce type I error, given the various dependent variables were conceptually related to each other. Movement irregularity was compared for each quartile using paired t-tests or Wilcoxon signed rank test when normality could not be assumed. A comparison between groups for frequency of responders was determined using the Chi-squared test.

#### 4.2.4 Results

T-tests revealed significant differences in movement evoked pain following the analgesia for all movements across both groups (table 4.2.3). Analgesia choices were similar amongst the groups with the exception of the ALBP group favouring a combination of analgesia and anti-inflammatory medication (figure 4.2.1). There were no significant differences between the groups for the amount of evoked pain experienced, with the exception of the lifting where the CLBP group reported greater pain (table 4.2.3).

Table 4.2.3. Degree of evoked pain before and after analgesia (VAS measured in mm).

	ALBP Pre	ALBP Post	t-test	CLBP Pre	CLBP Post	t-test
Flexion	41.1 (24.4)	20.7 (26.4)	0.003	39.6 (20.4)	3.0 (9.5)	<0.001
Extension	41.2 (19.6)	17.6 (19.8)	<0.001	40.9 (24.2)	17.2 (25.0)	<0.001
Side-Bending	40.4 (23.9)	18.3 (22.8)	<0.001	35.2 (21.9)	10.0 (22.3)	<0.001
Rotation	38.5 (21.2)	9.7 (17.7)	<0.001	37.0 (21.9)	7.8 (22.3)	<0.001
Lifting	31.6 (17.8)	15.2 (22.0)	0.001	52.2 (25.0)	28.3 (27.0)	<0.001

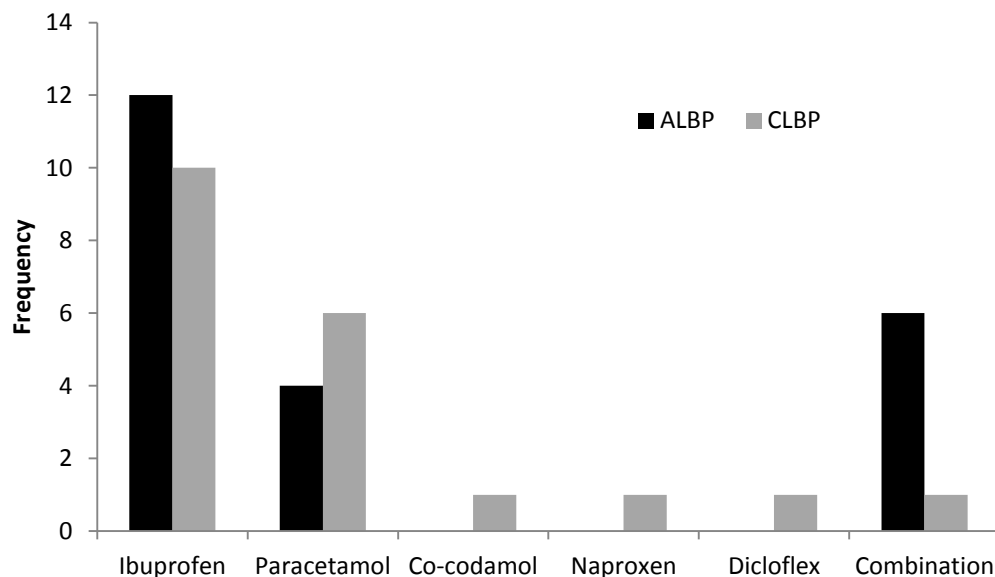


Figure 4.2.1. Medication choices by participants. Nb. Combination, Co-Codamol and Naproxen



The group kinematics prior to pain relief are presented in figure 4.2.2 and kinematic change following pain relief is presented in table 4.2.4.

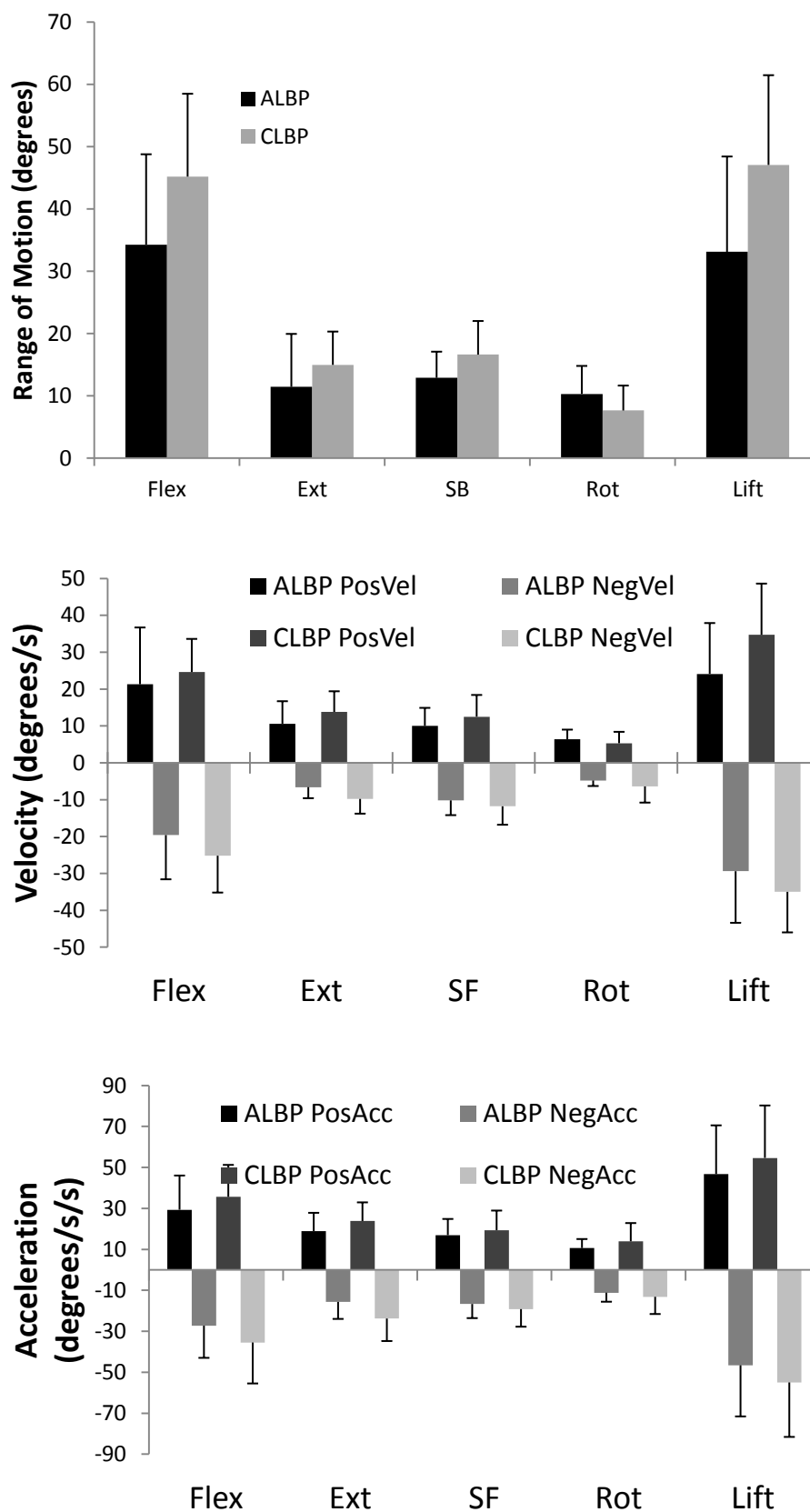


Figure 4.2.2. Lumbar kinematics of each group prior to pain relief

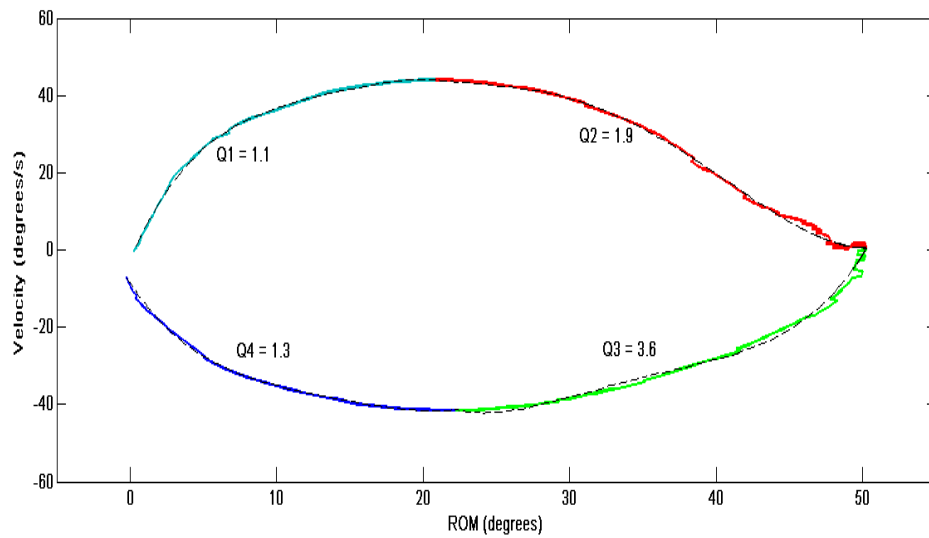
The results of the MANOVA show that there was no interaction between group and time ( $F = 0.114$ ,  $p = 0.989$ ). It was also demonstrated that time had no effect on kinematics ( $F = 1.234$ ,  $p = 0.293$ ) and that group had no effect on kinematics ( $F = 1.675$ ,  $p = 0.140$ ). These results therefore suggest that neither pain relief nor chronicity of pain had any effect on lumbar kinematics. As no significant effect was determined post hoc testing was unnecessary.

Pain relief had no effect on movement irregularity for each quartile of any movement tested (figure 4.2.3), with the exception of quartile 3 for flexion in the ALBP group, which demonstrated a reduction in movement irregularity (mean difference 1.1,  $p = 0.022$ ), and quartile 3 and 1 for side bending and rotation respectively in the CLBP group demonstrating an increase in movement irregularity (mean difference 1.2,  $p = 0.046$  side bending; mean difference 7.0,  $p = 0.024$  rotation).

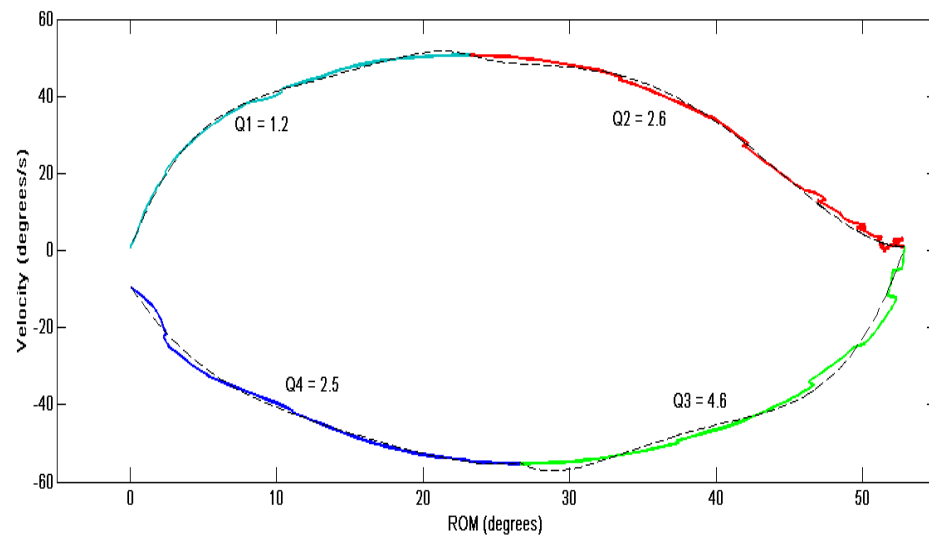
Table 4.2.4. Change in kinematic variables (mean (sd)) in response to pain relief. Nb. A positive value represents an increase in the specific variable, negative a reduction.

Flexion	ALBP	CLBP
Mean change in ROM ( $^{\circ}$ )	0.1 (4.3)	0.1 (6.6)
Mean change in Positive Velocity ( $^{\circ}\text{s}^{-1}$ )	0.3 (5.2)	2.6 (4.4)
Mean change Negative Velocity ( $^{\circ}\text{s}^{-1}$ )	4.8 (4.7)	1.1 (6.1)
Mean change Positive Acceleration ( $^{\circ}\text{s}^{-2}$ )	6.1 (12.3)	7.1 (8.1)
Mean change Negative Acceleration ( $^{\circ}\text{s}^{-2}$ )	6.9 (11.7)	0.5 (7.9)
Extension		
Mean change in ROM ( $^{\circ}$ )	-0.3 (2.2)	1.0 (2.8)
Mean change in Positive Velocity ( $^{\circ}\text{s}^{-1}$ )	0.3 (4.4)	1.8 (4.0)
Mean change Negative Velocity ( $^{\circ}\text{s}^{-1}$ )	0.6 (1.9)	2.1 (4.8)
Mean change Positive Acceleration ( $^{\circ}\text{s}^{-2}$ )	1.7 (9.6)	4.8 (8.4)
Mean change Negative Acceleration ( $^{\circ}\text{s}^{-2}$ )	1.8 (9.2)	5.1 (8.0)
Side-bending		
Mean change in ROM ( $^{\circ}$ )	-0.3 (1.6)	1.0 (2.0)
Mean change in Positive Velocity ( $^{\circ}\text{s}^{-1}$ )	1.1 (3.2)	2.1 (3.0)
Mean change Negative Velocity ( $^{\circ}\text{s}^{-1}$ )	0.7 (2.9)	2.7 (3.3)
Mean change Positive Acceleration ( $^{\circ}\text{s}^{-2}$ )	2.1 (7.5)	6.7 (7.1)
Mean change Negative Acceleration ( $^{\circ}\text{s}^{-2}$ )	3.1 (8.6)	6.9 (9.6)
Rotation		
Mean change in ROM ( $^{\circ}$ )	0.8 (2.0)	-0.2 (2.9)
Mean change in Positive Velocity ( $^{\circ}\text{s}^{-1}$ )	0.2 (2.0)	0.0 (2.1)
Mean change Negative Velocity ( $^{\circ}\text{s}^{-1}$ )	1.2 (1.7)	-0.7 (3.1)
Mean change Positive Acceleration ( $^{\circ}\text{s}^{-2}$ )	1.8 (4.8)	-1.4 (7.8)
Mean change Negative Acceleration ( $^{\circ}\text{s}^{-2}$ )	1.1 (4.4)	-0.7 (6.8)
Lifting		
Mean change in ROM ( $^{\circ}$ )	0.2 (5.5)	0.4 (4.5)
Mean change in Positive Velocity ( $^{\circ}\text{s}^{-1}$ )	1.5 (5.7)	-0.2 (9.7)
Mean change Negative Velocity ( $^{\circ}\text{s}^{-1}$ )	0.3 (5.2)	-0.6 (4.8)
Mean change Positive Acceleration ( $^{\circ}\text{s}^{-2}$ )	3.0 (10.9)	3.2 (16.3)
Mean change Negative Acceleration ( $^{\circ}\text{s}^{-2}$ )	0.0 (13.0)	3.3 (12.6)

ROM, range of motion; ALBP, acute low back pain; CLBP, chronic low back pain.



(a) Flexion trial VAS = 24



(b) Flexion trial VAS = 0

Figure 4.2.3. Spatial plots annotated with movement irregularity scores of a single participant for the movement of flexion. No effect of pain relief on movement irregularity is demonstrated.

The frequency of responders to pain relief for each movement in each group was determined, where responders were defined as those individuals who increased the kinematic variables by more than  $3^{\circ}$ ,  $4^{\circ}\text{s}^{-1}$ ,  $8^{\circ}\text{s}^{-2}$  for flexion and lifting;  $2^{\circ}$ ,  $3^{\circ}\text{s}^{-1}$ ,  $6^{\circ}\text{s}^{-2}$  for extension and side bending and  $1^{\circ}$ ,  $2^{\circ}\text{s}^{-1}$ ,  $4^{\circ}\text{s}^{-2}$  for rotation in response to pain relief. These values were selected with reference to the natural variation of repeated movements (chapter 4.1) (Williams et al., In

Press) and are believed to represent a clinically meaningful change in kinematics. Chi-squared testing revealed no significant difference in the number of responders between the groups for ROM ( $\chi^2 = 2.831$ ,  $p = 0.092$ ), angular velocity ( $\chi^2 = 0.012-1.878$ ,  $p = 0.171-0.913$ ), angular acceleration ( $\chi^2 = 0.001-0.689$ ,  $p = 0.406-0.972$ ) or movement irregularity ( $\chi^2 = 0.025-3.6$ ,  $p = 0.058-0.875$ ).

#### **4.2.5 Discussion**

The present study investigated the effects of targeted pain relief on lumbar kinematics in acute and chronic LBP sufferers. The results showed that the simple pain relief method was effective in reducing the magnitude of movement evoked pain by a clinically significant amount (Ostelo et al 2008) and are in-line with the only previous study reporting actual numbers for pain relief and kinematic change (Lilius et al., 1989). Whilst the reduction in pain was significant it is evident that both groups still reported levels of evoked pain. As the initial levels of evoked pain were similar for the movements tested this finding suggests that these medicines were only partially effective in relieving movement evoked pain. This may be due to the single one-off dose as all individuals avoided analgesia during the day of testing or that the strength of these medicines is not sufficient to fully relieve movement evoked LBP. Due to the partial pain relief, it is not clear whether the abolition of pain entirely would result in different findings. This finding may be important in light of the CSAG guidelines for the use of simple analgesia in the management of ALBP.

The findings of the present study demonstrate that partial pain relief, as achieved in this study, does not influence lumbar kinematics in either acute or chronic LBP. The findings are consistent with some of those within the literature and in contrast to others. In a CLBP sample, Jarzem et al., (2005) demonstrated no gains in flexion, extension and rotation following facet joint injection induced pain relief. Moreover, Davis and Kotowski (2005), also in CLBP sufferers

reported no significant gains in all ranges of motion following pain relief, which are both in agreement with the current study. However our findings, and those of the above studies are in contrast to other studies (Davis et al., 2005; Simmonds et al., 2003). Davis and Kotowski (2005) further reported a significant increase in lateral bending angular velocity and angular acceleration as well as rotation angular velocity and angular acceleration and Simmonds and Rebelo (2003) demonstrated a significant increase in sit-to-stand angular velocity following pain relief. These contrasts may be explained by the movement protocol, which utilised movement completed as fast as possible (Davis et al., 2005) rather than self selected, as in this study or by the pain relief method used, which was superficial heating and may therefore influence factors other than pain such as the compliance of musculoskeletal tissues (Bass et al 2007, Mutungi et al 1998). These differences highlight the unique methodology used in the current study – simple oral pain relief, which avoids soft tissue changes likely to be associated with other pain relief, thereby providing an isolated effect on pain. Improvements noted in other studies may be the result of biomechanical or neural changes in response to the chosen non-specific pain relief method, as the results of this study suggest that pain relief itself does not alter kinematics.

The results demonstrate that partial pain relief had no effect on the degree of movement irregularity as displayed though the use of spatial plots. The use of spatial plot analysis has been little studied in relation to LBP sufferers and these results are the first to investigate the effect of pain relief on movement irregularity. It appears that clinicians should not expect an immediate reduction in movement irregularity following interventions resulting in partial pain relief.

The current study found variability in the response to pain-relief, however group comparisons of the frequency of these responders did not show a significant difference between ALBP and CLBP. These results suggest that individuals with ALBP or CLBP are just as likely to respond to

pain relief and that neither group is more likely to be more 'sensitive' to pain relief. There was a trend towards the ALBP group demonstrating a greater ratio of responders to non-responders for the measurement of movement irregularity especially within quartiles 2 and 3 of the movement trajectory compared to the CLBP group, however this did not reach significance.

The findings of the current study contradict the many studies suggesting that trunk kinematics are reduced as a compensatory mechanism to avoid the provocation of pain (Shum et al., 2005a; 2005b; 2007a). If this were the case then it might be expected that a reduction of pain would result in automatic gains in kinematics, which was not observed in the current study. This suggests that perhaps the reduction in kinematics observed in LBP sufferers serves an alternative purpose or that the degree of pain relief was insufficient to cause an alteration in movement behaviour. It could represent a CNS shift to protect damaged tissue from further damage by providing 'more time' to process proprioceptive information, rather than the simple protection from evoked pain. Furthermore it may be reflective of deconditioning, which is commonly observed in CLBP sufferers (Verbunt et al., 2003), where the efficiency of the trunk is reduced, resulting in a reduction in kinematics (Shum, Crosbie, & Lee, 2009). It is however unclear if such deconditioning is evident in ALBP sufferers also.

The results of the current study may reflect the need for the 'system' to experience less or no pain for longer in order to cause an alteration in trunk function. The current study observed the immediate effects of pain relief and it is not clear if the same results would be replicated if the pain relief was continued over a period of time. This learning effect was observed by Moseley and Hodges (2005) in relation to induced pain and muscle function where repeated painful stimulus over seventy repetitions slowly resulted in changes in trunk muscle function and its gradual reverse following removal of the pain stimulus. The current study utilised only

three movement trials and there was no indication of pattern between the first to the third trial, however alteration over a longer time frame is possible.

The current study made the assumption that the LBP sufferers would display altered kinematics during spinal movement as this is commonly displayed in the literature (Marras et al., 2001; Marras et al., 1999; Marras & Wongsam, 1986; Shum et al., 2005a; 2005b; Wong et al., 2004). The resulting kinematic profiles demonstrated by the participants in this study illustrate that this assumption was valid, as direct comparisons to large normative databases demonstrates large differences in kinematic values (van Herp et al., 2000).

The results of this study are limited to the immediate effects of pain relief in an attempt to remove potential confounding variables that might influence the results. This meant the effects of analgesia on kinematics over time were not investigated. The sample size was limited and constrained to those with LBP confined to the back, therefore extrapolation to other types of LBP is not possible. The current study was limited to spinal movements and the findings might not be applicable to other tasks.

#### **4.2.6 Conclusion**

This study showed that partial pain-relief did not directly lead to changes in lumbar kinematics in ALBP or CLBP. Simple pain relief also had little effect on movement irregularity. Variability in response was observed, however the frequency of positive response was no different for ALBP or CLBP sufferers. Clinicians should not expect an immediate attenuation of lumbar kinematics following interventions which achieve partial pain relief.



### 4.3 Summary of results and key findings

The results of this chapter demonstrate that inertial sensors are a viable option for the objective measurement of lumbar kinematics including the higher order kinematics of angular velocity and angular acceleration. This has now been demonstrated in LBP suffers within a clinic based environment. The measurement method was shown to offer good repeated measures reliability for kinematic patterns (CMC) and peak kinematics (ICC). Furthermore small RMSEs and absolute mean differences were detected. These errors were small enough to make this system viable for detecting clinically meaning changes in lumbar kinematics.

The additional temporal kinematics obtained can be used to investigate the spatial relationship between these variables. Plots of movement against angular velocity provide a visualisation of movement trajectory and in doing so offer the opportunity to observe the coordination between these variables and therefore provide a measure of movement irregularity. This work demonstrates this process and offers a method of quantification of movement irregularity enabling the user/clinician to observe, quantify and compare the degree of movement irregularity across different segments of the movement. The results show this method to be reliable and also provide data to suggest the natural variation in this measurement across repeated movement trials. The comparison across quartiles demonstrated that the second quartile resulted in the greatest score of movement irregularity which corresponds to the individual approaching the end of range of the movement. Further analysis demonstrated little difference between ALBP and CLBP in measurement of movement irregularity with the exception of extension showing greater irregularity for the ALBP group.

The results demonstrated that there was no difference in ALBP and CLBP kinematics and that the partial pain relief achieved in this study had no impact on lumbar kinematics. It was also demonstrated that on the whole partial pain relief did not affect movement irregularity. This suggests that automatic attenuation of lumbar kinematics do not occur with partial pain relief

and therefore clinicians should not expect changes in lumbar kinematics following therapeutic interventions achieving partial pain relief.

Comparison of the frequency of response between the ALBP and CLBP group demonstrated that neither group was more likely to respond to pain relief. This demonstrates that neither group was more 'sensitive' to pain relief as measured by frequency for response.

#### Summary of key findings

- The inertial sensor system is a reliable method of obtaining lumbar kinematic measurements in LBP sufferers within a clinic environment, including angular velocity and angular acceleration.
- Spatial plot analysis can be use to quantify movement irregularity.
- Most commonly the second quartile of the movement (towards end range) demonstrates the greatest movement irregularity.
- ALBP sufferers demonstrate greater irregularity during the second quartile of extension compared to the CLBP group.
- There was no difference in kinematic variables between the groups.
- Partial pain relief had no effect on lumbar kinematics including movement irregularity.
- The number of responders was not significantly different between the ALBP and CLBP group suggesting that neither ALBP nor CLBP sufferers were more likely to respond to pain relief.

# **Chapter 5:**

## **Lumbar electromyography**

## **5.1 The effects of pain relief on the lumbar multifidus and iliocostalis EMG.**

### **5.1.1 Introduction**

It has been well documented that LBP sufferers display alterations in their trunk muscle activities (Geisser et al., 2005; Hodges et al., 2003b; van Dieen et al., 2003; Williams et al., 2010b). Alterations have been described in LBP sufferers in relation to the timing of muscle onset (Leinonen et al., 2001), peak amplitude of muscle activity (Sihvonen, Lindgren, Airaksinen, & Manninen, 1997) as well as changes in the overall profile of the EMG signal (Dankaerts et al., 2007). Despite these observations the underlying mechanism behind such alterations is not well understood.

It has been suggested that muscle function changes may be driven by pain (Hodges et al., 2003b). In order to test this hypothesis experimentally induce pain models have been used (Arendt-Nielsen et al., 1995; Dickx et al., 2008; Hodges et al., 2003a; Kiesel et al., 2008; Lamothe et al., 2004; Moe-Nilssen et al., 1999; Moseley et al., 2005; Moseley et al., 2004; Zedka et al., 1999) where temporary LBP is 'created' in previously healthy individuals. These studies demonstrate that induced pain can result in a delayed onset of transversus abdominus and an earlier onset of multifidus (Hodges et al., 2003a) or no effect on multifidus (Moseley et al., 2004). Furthermore an increase in peak EMG amplitude (Arendt-Nielsen et al., 1995; Lamothe et al., 2004) of the erector spinae muscles has been shown, however others demonstrate findings consistent with reduced peak activation (Dickx et al., 2008; Kiesel et al., 2008; Zedka et al., 1999). Also it has been shown that induced pain caused an alteration in the EMG profile by way of the loss of the flexion relation response (Zedka et al., 1999). Despite these findings experimentally induced pain fails to closely mimic clinical pain as the pain is constant in nature with little deviation other than gradual reduction over time and is likely to be due to chemical irritation of nociceptors within the muscle, the presence of which in LBP is not known, therefore these findings may not truly reflect the clinical situation (Williams et al., 2010b).

In order to overcome these limitations some authors have utilised an experimental pain relief model to study the effects of pain on trunk muscle function (Holm et al., 2000; Rashid et al., 2003). These studies show that pain relief resulted in an increase in Sorenson test performance (Rashid et al., 2003) but not in isokinetic dynamometry (Holm et al., 2000), however no EMG measurements were conducted meaning the effects of pain relief on EMG related muscle functions are not clear.

Therefore it remains unclear whether pain relief causes an alteration in trunk muscle function as measured by EMG. LBP management strategies often target pain relief as a method of altering the function of the trunk (Jette et al., 1994) however before management can be rationalised it needs to be determined as the effect of pain relief on trunk muscle functions.

### **5.1.2 Aim of study**

The aim of the current study was to investigate the effects of pain relief on lumbar trunk muscle function in ALBP and CLBP sufferers.

### **5.1.3 Methods**

#### **5.1.3.1 Participants**

Forty volunteers were recruited from local physiotherapy and chiropractic clinics. Participants were screened by a physiotherapist for inclusion and exclusion criteria (table 5.1.1) and were divided into one of two groups according to duration of pain (table 5.1.2). There were no significant differences between the groups except for duration and severity of pain. The severity of the pain was rated using a visual analogue scale (0 to 100mm) where participants were asked to mark their average pain over the preceding week, and fear of movement was evaluated using the Tampa scale of kinesiophobia questionnaire.

All participants were supplied with information sheets and gave informed consent. The study was approved by the Research Ethics Committee of the National Health Service (08/H1111/38).

Table 5.1.1. Inclusion and exclusion criteria

Inclusion	Exclusion
Pain confined to between lower ribs and inferior gluteal folds	History of tumors
Movement evoked pain	Spinal fractures
Aged 18-55 years old	Surgery
Seeking healthcare for LBP	Neurological signs or symptoms
Acute – Pain present for less than 3 weeks on a history of no pain for at least 12 months	Rheumatological or Neurological disease
Chronic – Pain present on at least 3 days per week for at least 52 weeks	Known spinal deformity

Table 5.1.2. Participant demographics.

	ALBP	CLBP	P-value (t-test)
Male/Female	11/9	11/9	
Age (years)	42.7 (6.8)	36.6 (10.8)	0.0754
Height (m)	172.9 (11.3)	173.6 (11.2)	0.6994
Weight (kg)	82.6 (16.6)	83.7 (16.1)	0.5467
BMI (Kg/m <sup>2</sup> )	27.5 (4.0)	26.2 (4.1)	0.4337
TSK	39.0 (4.8)	38.9 (6.9)	0.8541
VAS (mm)	62.2 (16.6)	46 (22)	0.018
Duration	12.3 (6.7) days	9.4 (7.4) years	<0.0001

\* p < 0.05; BMI, body mass index; TSK, tampa scale of kinesiophobia; VAS, visual analogue scale.

#### 5.1.3.2 Procedure

Participants stood upright with feet positioned naturally along a line before completing three trials of flexion, extension, left side bending, right side bending, left rotation, right rotation and

a box lift. The box was positioned on markers ensuring identical placement and measured 460 x 260 x 300mm and weighed 3kg. The magnitude of the worst pain evoked by the movements was measured using a visual analogue scale completed following each of the movements. Participants were then requested to self-administer their usual oral analgesia and given a break of between 45-60 minutes after which the movements were repeated. There were no restrictions on the type of analgesia used and the sensors remained attached throughout the experiment.

#### 5.1.3.3 Instrumentation

Bipolar EMG electrodes, with a fixed electrode distance of 20mm, were attached to the skin. No skin preparation was carried out, as recommended by the company due to the typical Input Impedance of >10,000,000 MOhms. EMG was collected using a portable EMG system (PS850, Biometrics Ltd. UK) with a bandwidth of 20-450Hz and a common mode rejection ratio of >96dB (typically 110dB) at 60Hz. All raw signals were pre-amplified with a gain of 1000 and data were captured at 1000Hz and transferred to a portable datalogger for later analysis. Two electrodes were placed bilaterally over the lumbar multifidus (2-3cm lateral of the L5 spinous process on a line intersecting the L1-2 interspinous space and caudal tip of the posterior superior iliac spine) and lumbar iliocostalis muscles (1-2cm medial from the line from the posterior superior iliac spine to the lowest point of the lower rib, at the level of L2) as recommended by SENIAMS (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). Therefore the left lumbar multifidus (LLM), right lumbar multifidus (RLM), left iliocostalis (LIC) and right iliocostalis (RIC) were investigated in this study. The reference electrode was strapped over the right ulnar styloid process (Thomas & Lee, 2000). Two inertial sensors (Microstrain, VT, USA) were synchronised with the EMG system and used to measure lumbar kinematics with one placed over the S1 spinous process and one over the L1 spinous process (figure 5.1.1). Lumbar spine movements were calculated from the relative orientations between the two sensors

computed from the direction cosine matrices ((Burnett et al., 1998; Lee et al., 2003; Williams et al., In Press).

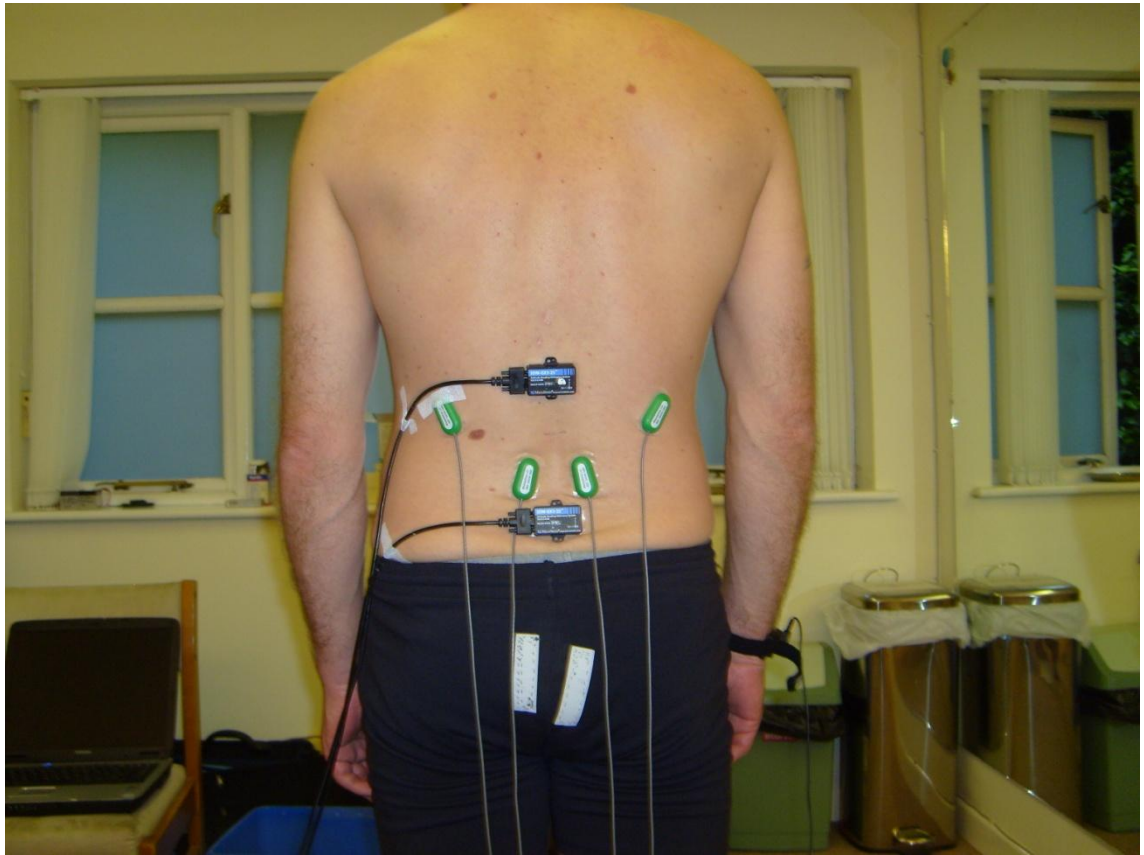


Figure 5.1.1. Experimental set up.

#### 5.1.3.4 Data analysis

The raw data were transferred to Matlab (Mathworks R2008b) for processing. Raw data were demeaned; full-wave rectified and low pass filtered using a zero-lag 4<sup>th</sup> order low pass butterworth filter at 5Hz to create a linear envelope for each channel. Repeated measures reliability were determined by separating the data into three movement cycles, time-normalising and then calculating the coefficient of multiple correlation (CMC) and intraclass correlation coefficient (ICC). Reliability was calculated by comparing the time-normalised EMG linear envelopes for each of the three repeated movement cycles for each movement investigated. CMC computes a measure of similarity between the EMG linear envelopes for repeated movement trials and the ICC was used to determine the reliability of peak EMG amplitude measurements.



Linear envelopes were inspected for overall profile and profile patterns were determined according to the specific movements. EMG profile patterns were defined as demonstrating either an increase in EMG activity, decrease in EMG activity or no defined pattern in EMG activity as seen during the movement. These were determined relative to the deviation of the EMG signal by greater than  $\pm 3$  standard deviations of the mean during standing. The classified linear envelopes were cross-correlated to the displacement data for each of the movements to yield the peak correlation coefficient and the time-lag between the two signals. Cross-correlation analyses the degree of correlation between two wave-forms over time by sequentially shifting one of the wave-forms forwards and backwards to determine the point of maximum correlation, when the wave forms are most in-phase. The amount of shift necessary represents the lag between the two wave-forms and represents the muscle onset relative to the displacement (Kuriki, Micolis de Azevedo, Filho, & Alves, 2011). The onset was calculated by comparing the entire EMG data train (three cycles of movement) with the whole displacement data train (three displacement cycles). Due to the flat nature of the profile with no defined patterned, cross correlation was only completed on the defined profiles classified as increase or decrease EMG profiles.

The differences between groups were determined using unpaired t-tests (or Mann Whitney U when normality could not be assumed) and the effects of pain relief were determined using paired t-tests (or Wilcoxon sign rank test when normality could not be assumed), for muscle onset and peak magnitude.

#### **5.1.4 Results**

##### **5.1.4.1 Reliability**

The overall mean (sd) coefficient of multiple correlation (CMC) for the EMG envelopes during repeated trials were found to be 0.68 (0.01) for the ALBP group and 0.71 (0.01) for the CLBP, suggesting that the sequential EMG profiles were similar in shape. The mean intraclass

correlation coefficients for the peak magnitude of EMG activity were 0.97 (0.01) for the ALBP group and 0.96 (0.03) for the CLBP group indicating highly reliable peak magnitude data.

#### 5.1.4.2 Flexion

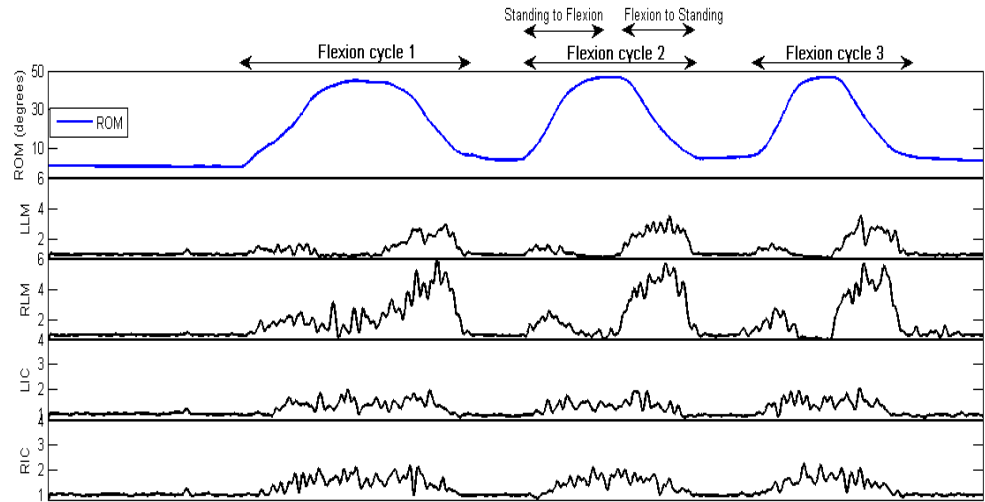
##### 5.1.4.2.1 Pattern

In those who reported evoked pain, different subjects appeared to show different EMG profiles during flexion, which generally fall into three types – (1) an increase in EMG activity profile (n=5 ALBP; n=8 CLBP), (2) a decrease in EMG activity profile (n=4 ALBP; n=1 CLBP) and (3) no pattern (n=5 ALBP; n=1 CLBP) (figure 5.1.2 (a), (b), (c)).

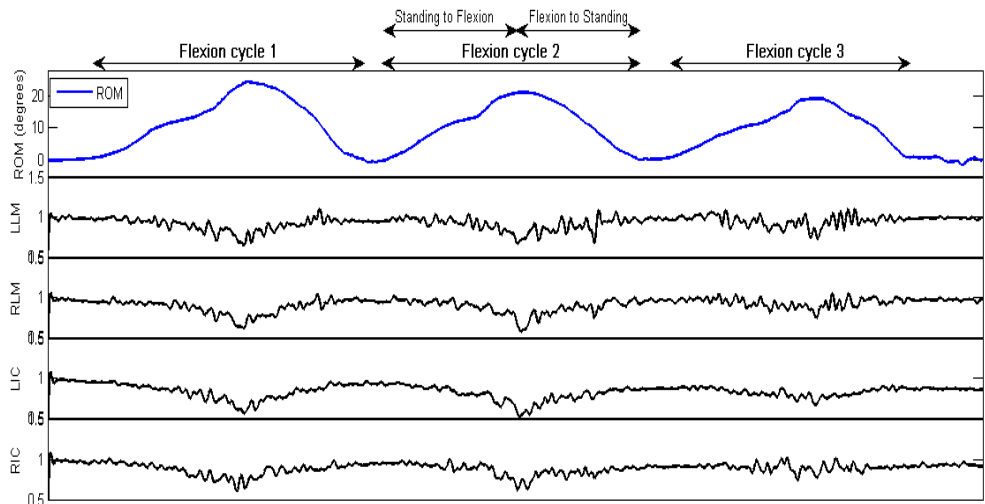
The increase in EMG activity profile shows the initial phase of motion occurs with an increase in activity of all muscles as they work to eccentrically control the trunk. As the individual approaches end of range flexion, lumbar multifidus undergoes a period of reduced activity which has been described as the flexion relaxation response. No such response is evident for the iliocostalis muscles. The return from flexion to standing is associated with a large spike in concentric lumbar multifidus activity and consistent iliocostalis activity. These concentric spikes are greater than the eccentric spikes for the lumbar multifidus muscles. Iliocostalis displays a fairly consistent level of activity throughout the movement.

The decrease in EMG activity profile illustrates poorly defined onset and eccentric spikes suggesting eccentric muscle levels similar to standing. A reduction in muscle activity occurs near end range and is evident in all muscles. No large concentric muscle spikes are evident on the return from flexion.

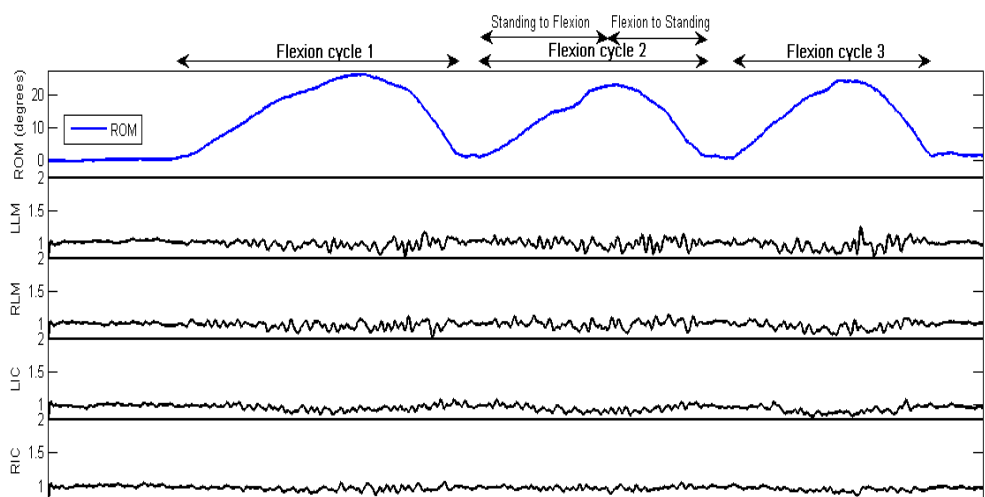
The no pattern profile shows little or no spikes or drops in muscle activity suggesting a continuous level of EMG activity is maintained throughout the movement.



(a) Increase EMG profile



(b) Decrease EMG profile



(c) No profile

Figure 5.1.2. EMG profiles evident during flexion trials with ROM at the top and EMG profiles below (a) Increase EMG profile; (b) Decrease EMG profile; (c) No pattern profile. LLM, left lumbar multifidus; RLM, right lumbar multifidus; LIC, left iliocostalis; RIC, right iliocostalis.

The cross-correlation coefficients were calculated for those EMG profiles demonstrating a definable pattern only, the increase in EMG pattern and decrease in EMG pattern. Cross correlations were high for the EMG activity profiles across both groups and classifications for the movement of flexion (table 5.1.3).

Table 5.1.3. Mean (sd) cross correlation between classified EMG profile and lumbar movement for the movement of flexion

Flexion	ALBP		CLBP	
	Increase EMG profile	Decrease EMG profile	Increase EMG profile	Decrease EMG profile
n (%)	5 (36%)	4 (29%)	8 (80%)	1 (10%)
Cross correlation				
LLM	0.81 (0.06)	0.74 (0.04)	0.76 (0.08)	0.66
RLM	0.81 (0.07)	0.74 (0.02)	0.76 (0.08)	0.66
LIC	0.83 (0.06)	0.75 (0.03)	0.78 (0.04)	0.65
RIC	0.81 (0.06)	0.75 (0.03)	0.79 (0.02)	0.65

It is evident that those individuals displaying no profile EMG pattern recorded significantly smaller ranges of motion, velocities and accelerations compared with the increase EMG profile ( $U = 0$ ,  $p < 0.001$  ROM,  $U = 1$ ,  $p = 0.001$  PosVel;  $U = 2.5$ ,  $p = 0.002$  NegVel;  $U = 6$ ,  $p = 0.005$  PosAcc;  $U = 6$ ,  $p = 0.005$ ) and significantly smaller ROM ( $U = 30$ ,  $p = 0.006$ ), negative velocity ( $U = 27$ ,  $p = 0.029$ ) and negative acceleration ( $U = 26.5$ ,  $p = 0.036$ ) compared with the decrease EMG profile (table 5.1.4).

Table 5.1.4. Mean (sd) kinematic values relating to specific EMG profiles.

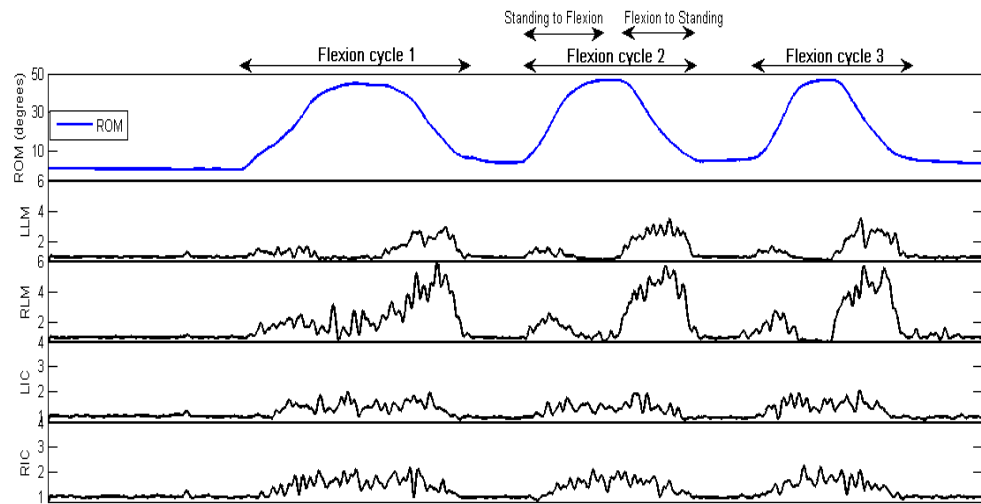
Kinematics	Double peak	Single drop	No Pattern
ROM ( $^{\circ}$ )	46 (10.5)*	43 (14.7)*	20 (5.2)
Pos Vel ( $^{\circ}\text{s}^{-1}$ )	28 (8.3)*	26 (22.7)	10 (3.9)
Neg Vel ( $^{\circ}\text{s}^{-1}$ )	26 (9.9)*	27 (16.8)*	12 (1.9)
Pos Acc ( $^{\circ}\text{s}^{-2}$ )	38 (12.7)*	35 (23.6)	17 (2.7)
Neg Acc ( $^{\circ}\text{s}^{-2}$ )	41 (20.9)*	32 (18.3)*	16 (2.4)

\*  $p < 0.05$  compared with kinematic values associated with the no pattern profile. ROM, range of motion; Pos Vel, positive velocity; Neg Vel, negative velocity; Pos Acc, positive acceleration; Neg Acc, negative acceleration.

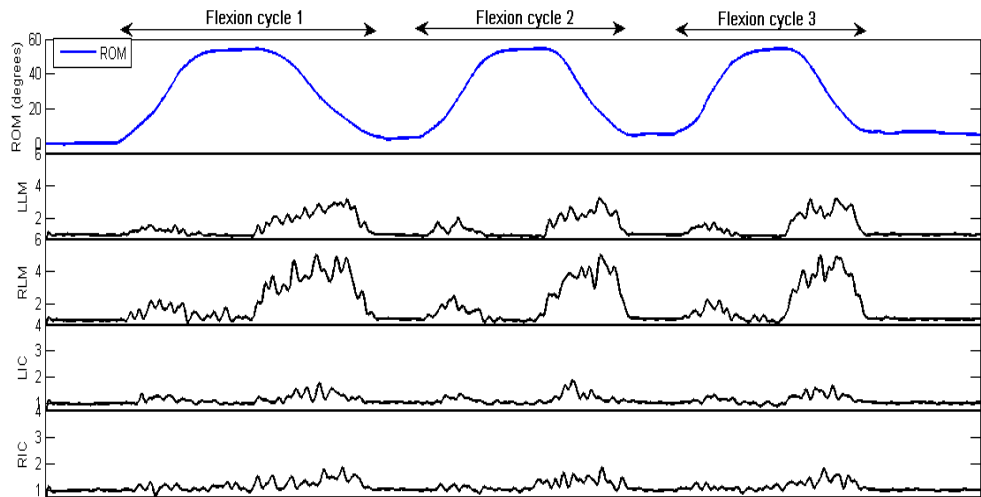
#### 5.1.4.1.2 Pain relief

Of the nine ALBP sufferers who reported pain relief, two changed their overall pattern. One switched from no pattern to a decrease EMG profile pattern and the other from the decrease EMG profile to no pattern. No other alterations in EMG profile were determined.

In the CLBP group no individuals altered the profile of EMG activity (figure 5.1.3).



(a) Pre pain relief flexion trials (VAS = 38).



(b) Post pain relief flexion trials (VAS 24).

Figure 5.1.3. Effect of pain relief on flexion EMG profile – no change demonstrated. (a) Participant flexion trials prior to (VAS = 38) and following (b) pain relief (VAS = 24).

EMG onset was not affected by pain relief in those demonstrating an increase EMG profile classification ( $z = 0.89$ ,  $p = 0.37$  for ALBP;  $z = 0.41$ ,  $p = 0.68$  for CLBP) or the decrease EMG profile ( $z = 0$ ,  $p = 1$  for ALBP) during flexion (table 5.1.5).

Table 5.1.5. Mean (sd) values of the muscle onset times (ms) for the increase EMG profile group and decrease EMG profile group for the movement of flexion (a negative value shows the muscle active following the onset of lumbar movement).

	Increase profile		Decrease profile	
	Pre (ms)	Post (ms)	Pre (ms)	Post (ms)
ALBP				
LLM	-36.8 (74.9)	-67.7 (33.5)	23.9 (82.7)	-15.6 (87.8)
RLM	-61.8 (56.4)	-66.9 (35.0)	50.0 (55.2)	14.4 (74.9)
LIC	-17.8 (29.7)	-44.7 (38.3)	28.1 (78.0)	4.1 (85.1)
RIC	-10.9 (7.4)	-37.5 (35.2)	-7.1 (61.2)	2.7 (77.6)
CLBP				
LLM	-7.4 (81.7)	-15.9 (84.6)	108.0	113.8
RLM	-41.6 (62.6)	-25.5 (86.9)	96.1	112.9
LIC	-20.1 (64.4)	-26.0 (27.8)	-72.8	103.7
RIC	-30.1 (55.5)	-16.7 (55.9)	-65.2	-62.5

Peak EMG activity was not significantly affected by pain relief in those who displayed the increased EMG profile ( $z = 0 - 0.89$ ,  $p = 0.37 - 1$  for ALBP;  $z = 0.76$ ,  $p = 0.45$  for CLBP) (figure 5.1.4) or the decrease EMG profile ( $z = 0$ ,  $p = 1$  for ALBP (figure 5.1.5) during flexion.

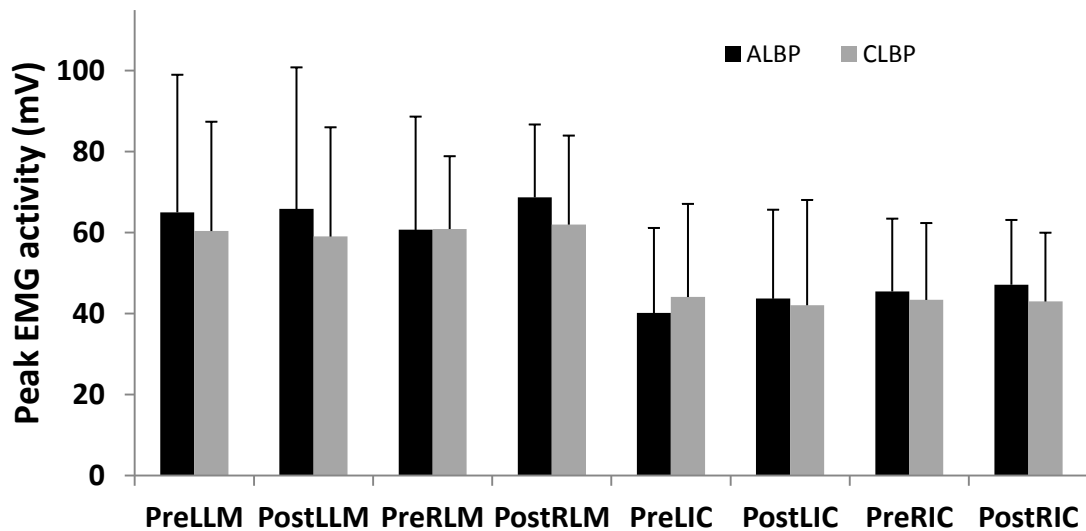


Figure 5.1.4. Mean pre and post pain relief peak EMG values for the increase EMG profile during flexion (error bars represent one standard deviation).

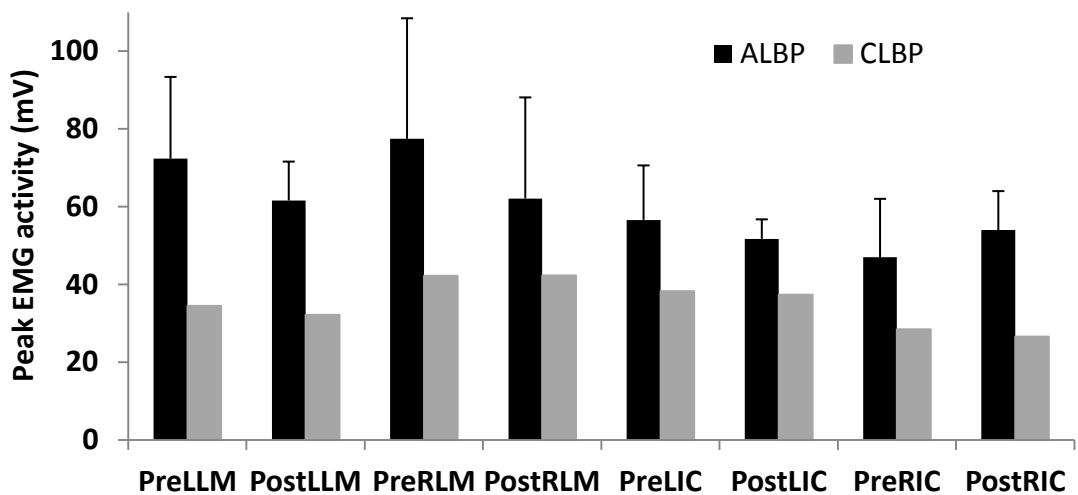


Figure 5.1.5 Mean pre and post pain relief peak EMG values for the decrease EMG profile during flexion (error bars represent one standard deviation, not evident for CLBP as  $n = 1$ ).

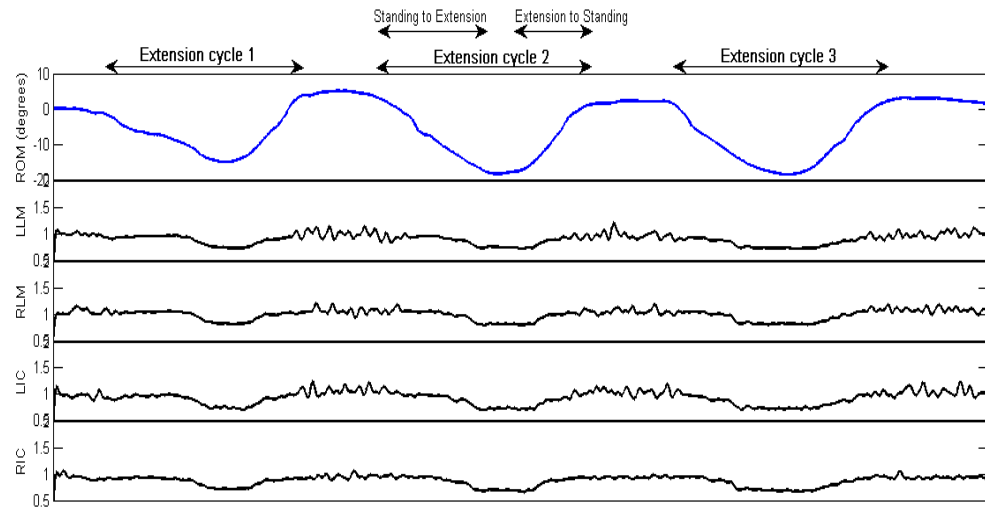


#### 5.1.4.3 Extension

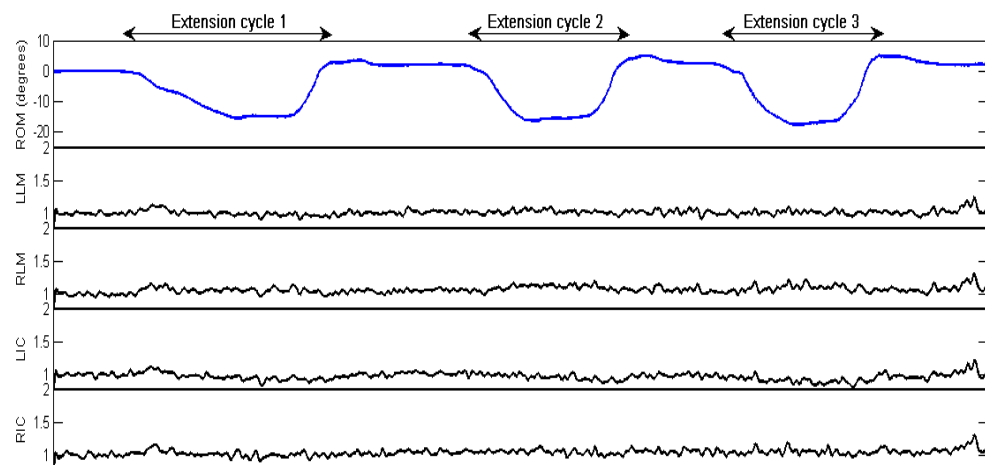
##### 5.1.4.3.1 Pattern

In those who reported evoked pain during extension, different subjects appeared to show different EMG profiles, which generally fall into two types – (1) a decrease in EMG activity profile (n = 4 ALBP; n = 5 CLBP) and (2) no pattern (n = 9 ALBP, n = 7 CLBP) (figure 5.1.6). The decrease EMG profile demonstrates a level of resting activity in standing and once motion begins the activity of all muscles drops slightly until mid range when a significant further drop is observed which flattens near end range. This region could be described as an extension relaxation response. Activity begins to increase where the individual is close to upright standing. The pattern is similar throughout all the muscles.

The envelope described as no pattern demonstrate no discernable peaks and troughs in the EMG signal suggesting a consistent level of muscle activity throughout the movement. No kinematic variables appeared to define the EMG patterns (U = 41 - 53.5, p = 0.316 – 0.867).



(a) Decrease EMG profile



(b) No profile

Figure 5.1.6. EMG profiles evident during extension trials with kinematics at the top and EMG profiles below (a) reduced EMG activity profile; (b) no profile. LLM, left lumbar multifidus; RLM, right lumbar multifidus; LIC, left iliocostalis; RIC, right iliocostalis

The cross-correlation coefficients were moderate to high for the decrease EMG activity profiles across both groups for the movement of extension (table 5.1.6).

Table 5.1.6. Mean (sd) cross correlation between EMG profile and lumbar extension movement.

	ALBP	CLBP
Extension	Decrease EMG profile	Decrease EMG profile
n (%)	4 (31%)	5 (42%)
Cross correlation		
LLM	0.58 (0.17)	0.70 (0.04)
RLM	0.58 (0.17)	0.69 (0.04)
LIC	0.58 (0.17)	0.72 (0.06)
RIC	0.58 (0.17)	0.71 (0.05)

#### 5.1.4.3.2 Pain relief

All ALBP and CLBP sufferers demonstrated no change in EMG pattern during extension following pain relief. Pain relief did not affect the muscle onset time for the ALBP ( $z = 0$ ,  $p = 1$ ) or CLBP ( $z = 0 - 0.89$ ,  $p = 0.371 - 1$ ) group.

#### 5.1.4.4 Left Side Flexion

##### 5.1.4.4.1 Pattern

In those who reported evoked pain during LSF, different subjects demonstrated different EMG profiles, which generally fall into two types – (1) the ipsilateral increase pattern ( $n=6$  ALBP;  $n=4$  CLBP) (figure 5.1.7) (described below) and (2) no pattern ( $n=3$  ALBP;  $n=5$  CLBP). The motion begins with very little additional initial muscle activity after which the left iliocostalis begins to become active. This activity spike continues, demonstrating the left iliocostalis is functioning concentrically to pull the trunk down. This activity is accompanied by a small amount of bilateral lumbar multifidus activity. The return to upright is associated with little activity in the left sided muscles and a small degree of activation in the right iliocostalis primarily.

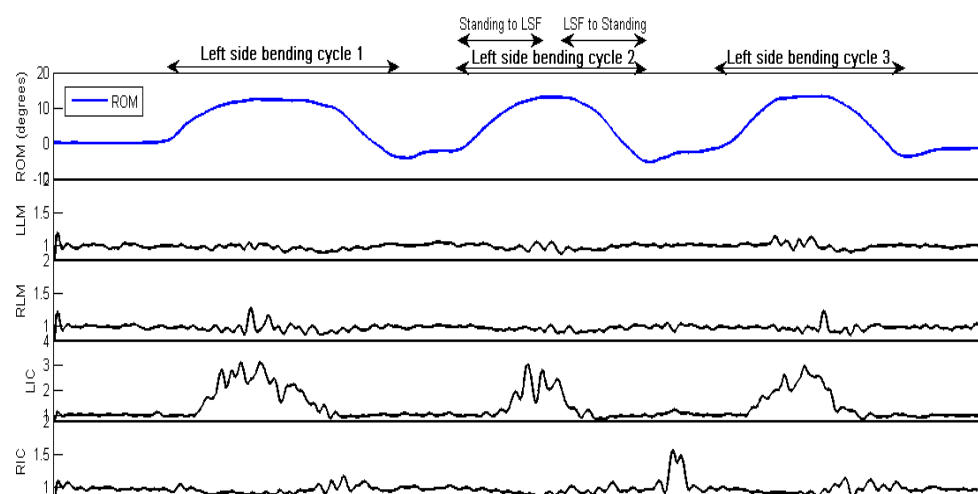


Figure 5.1.7. EMG profile during LSF – Increase ipsilateral profile with kinematics at the top and EMG profiles below. LLM, left lumbar multifidus; RLM, right lumbar multifidus; LIC, left iliocostalis; RIC, right iliocostalis

The cross-correlation coefficients were moderate to high for the ipsilateral increase EMG activity profile across both groups for the movement of left side flexion (table 5.1.7).

Table 5.1.7. Mean (sd) cross correlation between EMG profile and left side flexion movement.

	ALBP	CLBP
	Increase ipsilateral	Increase ipsilateral
Left side flexion	EMG profile	EMG profile
n (%)	6 (67%)	4 (44%)
Cross correlation		
LLM	0.68 (0.08)	0.70 (0.05)
RLM	0.68 (0.08)	0.75 (0.11)
LIC	0.74 (0.07)	0.70 (0.03)
RIC	0.68 (0.08)	0.74 (0.10)

Those individuals demonstrating no pattern during the left side bending movement had a tendency towards less ROM, velocity and acceleration for both ALBP and CLBP groups however this was demonstrated to be non significant ( $U = 2 - 6$ ,  $p = 0.07 - 0.439$ ) (table 5.1.8).

Table 5.1.8. Mean (sd) kinematic values relating to ipsilateral side bending EMG profiles.

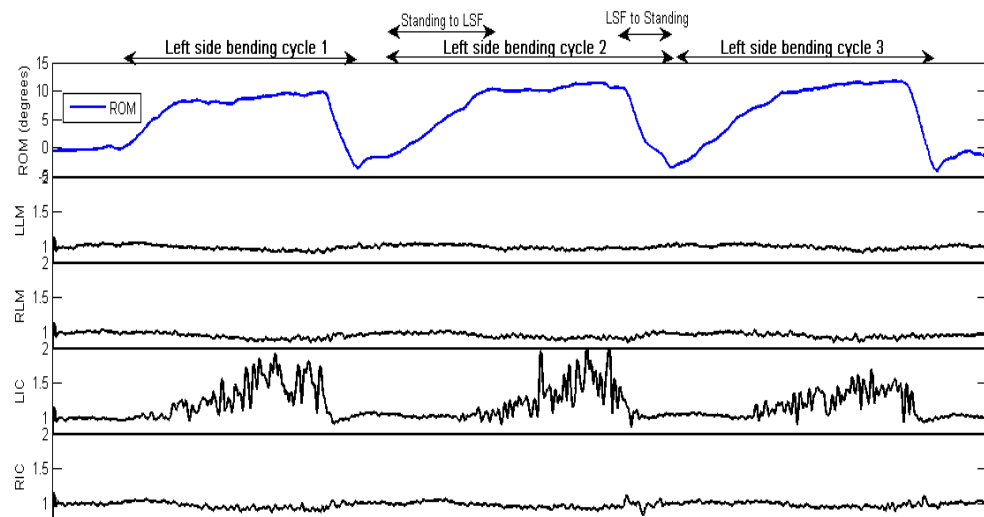
Kinematics	ALBP		CLBP	
	Ipsilateral increase	No profile	Ipsilateral increase	No profile
ROM ( $^{\circ}$ )	12 (1.7)	11 (6.0)	18 (5.2)	15 (7.0)
Pos Vel ( $^{\circ}\text{s}^{-1}$ )	10 (6.0)	6 (2.6)	13 (4.8)	11 (6.0)
Neg Vel ( $^{\circ}\text{s}^{-1}$ )	13 (0.5)	9 (4.3)	15 (4.2)	11 (6.1)
Pos Acc ( $^{\circ}\text{s}^{-2}$ )	17 (6.5)	10 (2.4)	21 (4.8)	19 (11.1)
Neg Acc ( $^{\circ}\text{s}^{-2}$ )	18 (3.3)	13 (4.0)	21 (7.4)	17 (9.1)
RSF				
ROM ( $^{\circ}$ )	18 (4.4)	15 (3.0)	19 (6.4)	16 (6.1)
Pos Vel ( $^{\circ}\text{s}^{-1}$ )	17 (3.2)*	10 (3.2)	18 (7.8)	12 (7.0)
Neg Vel ( $^{\circ}\text{s}^{-1}$ )	12 (5.0)	9 (2.0)	15 (6.5)	9 (4.5)
Pos Acc ( $^{\circ}\text{s}^{-2}$ )	25 (2.7)	15 (7.6)	30 (14.5)	17 (7.8)
Neg Acc ( $^{\circ}\text{s}^{-2}$ )	21 (8.5)*	13 (1.9)	24 (10.6)	17 (7.3)

\*  $p = <0.05$  compared with kinematic values associated with the no pattern profile. ROM, range of motion; Pos Vel, positive velocity; Neg Vel, negative velocity; Pos Acc, positive acceleration; Neg Acc, negative acceleration.

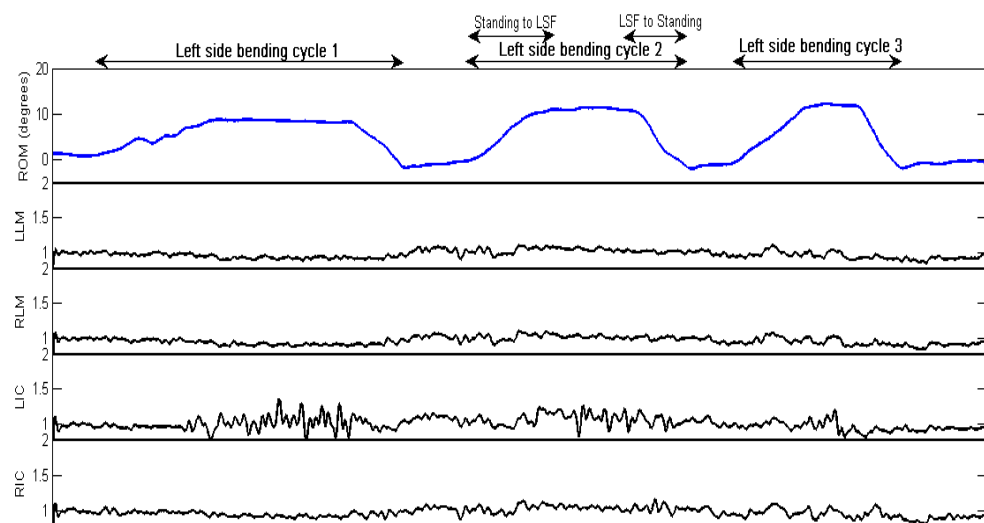
Two individuals (one from each group) demonstrated high levels of right lumbar multifidus activity which may represent a strategy to control the pull of gravity and the left iliocostalis muscle, with the effect of limiting lower lumbar movement into left side bending.

#### 5.1.4.4.2 Pain Relief

Two of the ALBP sufferers altered the pattern of their EMG profile in response to pain relief by flattening the linear envelope peak in the left iliocostalis (figure 5.1.8). This muscle, as described above, is operating to concentrically pull the trunk into LSF, therefore the reduction in activity may suggest that this movement is associated with less effort.



(a) Pre pain relief LSF trials (VAS = 20).



(b) Post pain relief LSF trials (VAS = 5).

Figure 5.1.8. Effect of pain relief on LSF EMG profile – flattening of the profile is demonstrated.

(a) Participant movement trials prior to (VAS = 20) and following (b) pain relief (VAS = 5).

EMG onset was not significantly affected by pain relief in those within the ipsilateral increase EMG profile during left side flexion for either ALBP ( $z = 0 - 1.22$ ,  $p = 0.221 - 0$ ) or CLBP ( $z = 0.5 - 1.5$ ,  $p = 0.134 - 0.6171$ ); however there was a consistent later activation compared to the movement following pain relief (table 5.1.9).

Table 5.1.9. Mean (sd) values of the muscle onset times (ms) for the ipsilateral increase EMG profile group for the movement of left side flexion (a negative value shows the muscle active following the onset of lumbar movement).

ALBP	Pre (ms)	Post (ms)
LLM	-57.5 (105.7)	-98.0 (75.9)
RLM	-52.0 (107.9)	-96.5 (75.8)
LIC	13.4 (35.0)	-40.1 (82.1)
RIC	-74.1 (71.4)	-100.6 (60.3)

CLBP	Pre (ms)	Post (ms)
LLM	15.9 (79.1)	-97.9 (143.6)
RLM	-26.3 (96.6)	-101.8 (58.2)
LIC	18.3 (31.5)	-18.1 (51.7)
RIC	33.2 (127.9)	-28.7 (153.7)

Peak EMG activity was not significantly affected by pain relief in those who displayed the ipsilateral increased EMG profile during left side flexion for the ALBP ( $z = 0 - 0.41$ ,  $p = 0.683 - 1$ ) or CLBP ( $z = 0 - 1.5$ ,  $p = 0.134 - 1$ ) group (figure 5.1.9).

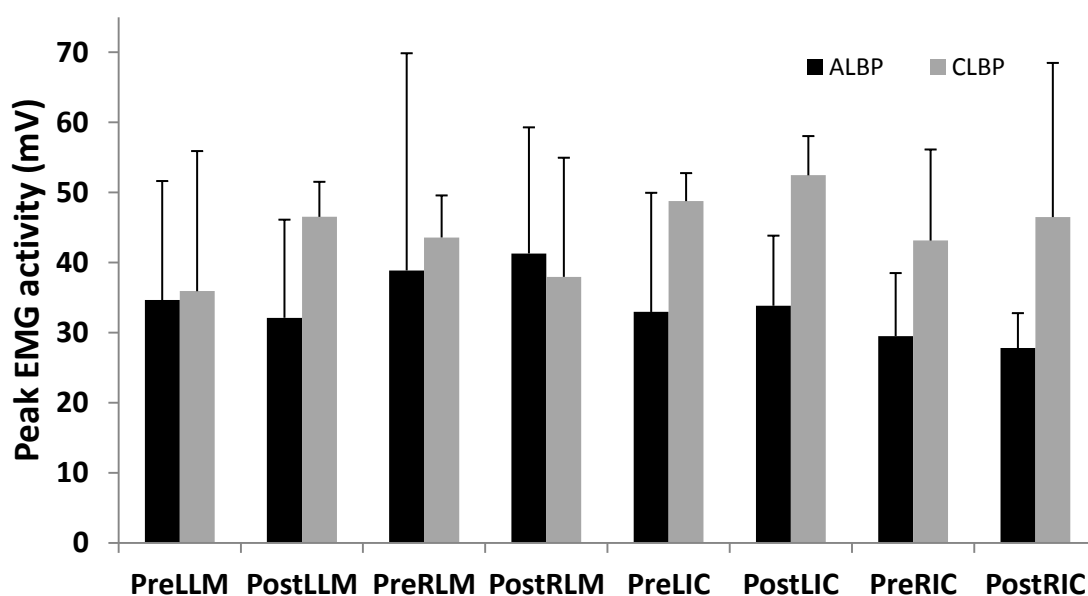


Figure 5.1.9. Mean pre and post pain relief peak EMG values for the ipsilateral increase EMG profile during left side flexion.

#### 5.1.4.5 Right Side Flexion

##### 5.1.4.5.1 Pattern

Right side flexion mirrors that of left side flexion where subjects demonstrated one of two patterns – (1) the ipsilateral increase EMG pattern, where the prime mover, RIC, operates concentrically to pull down the trunk (n=4 ALBP; n=3 LBP) or (2) no discernable pattern (n=6 ALBP; n=4 CLBP). Three individuals demonstrated activity in the LLM which as mentioned previously may be functioning to control the pull of RIC and gravity. The ipsilateral increase pattern demonstrated a trend towards greater kinematic values with significantly greater positive velocity ( $U = 24$ ,  $p = 0.011$ ) and negative acceleration ( $U = 23$ ,  $p = 0.019$ ) kinematic values observed for the ALBP group (table 5.1.7).

The cross-correlation coefficients were moderate to high for the ipsilateral increase EMG activity profile across both groups for the movement of RSF (table 5.1.10).

Table 5.1.10. Mean (sd) cross correlation between EMG profile and right side flexion movement.

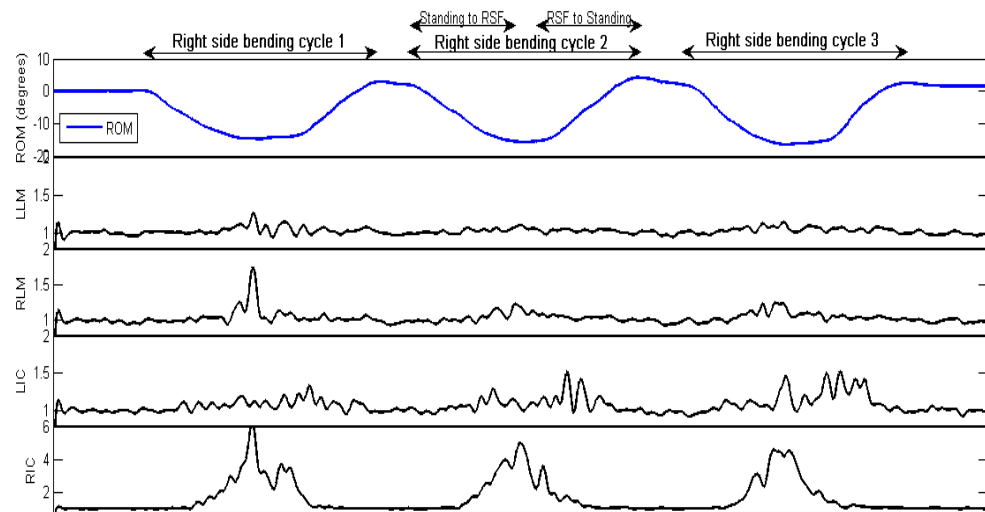
	ALBP	CLBP
Right side flexion	Increase ipsilateral EMG profile	Increase ipsilateral EMG profile
n (%)	4(40%)	4 (43%)
Cross correlation		
LLM	0.77 (0.10)	0.69 (0.13)
RLM	0.75 (0.07)	0.68 (0.13)
LIC	0.76 (0.06)	0.71 (0.16)
RIC	0.79 (0.04)	0.63 (0.04)



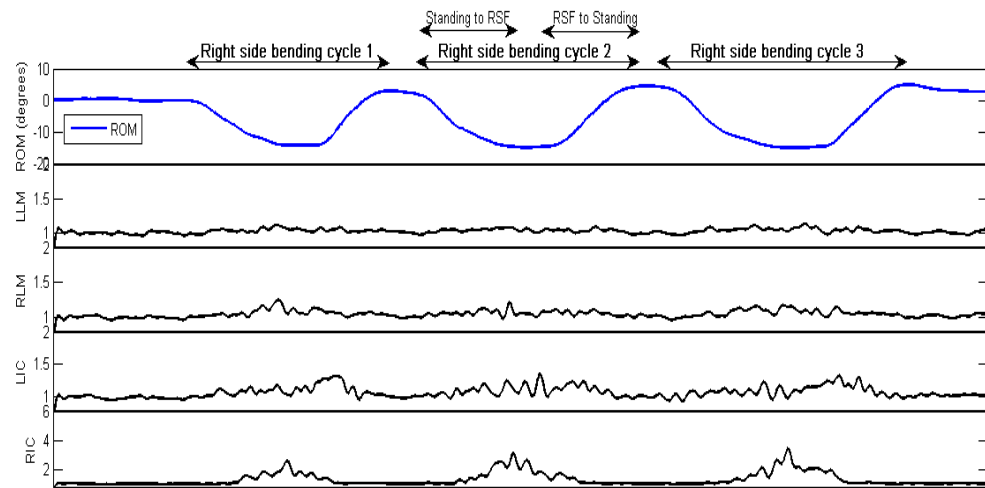
#### 5.1.4.5.2 Pain relief

Three of those individuals reporting pain relief in the ALBP group demonstrated an alteration in the EMG pattern where there was a flattening of the linear envelope for the right Iliocostalis demonstrating a shift from the ipsilateral increase pattern towards a no pattern profile (figure 5.1.10). No change in pattern was observed in the remaining individuals.

Four of the CLBP group changed their EMG profile however the change noted was varied. Two individuals demonstrate a shift from no pattern to an ipsilateral increase pattern, one shifted from an ipsilateral increase pattern to no pattern and one shifted from no pattern to a pattern which demonstrates the LIC was used to pull the trunk up (contralateral increase).



(a) RSF pre pain relief (VAS = 17).



(b) RSF post pain relief (VAS = 0).

Figure 5.1.10. Effect of pain relief on RSF EMG profile – flattening of the profile is demonstrated. (a) Participant movement trials prior to (VAS = 17) and following (b) pain relief (VAS = 0). LLM, left lumbar multifidus; RLM, right lumbar multifidus; LIC, left iliocostalis; RIC, right iliocostalis

EMG onset was not significantly affected by pain relief in those within the ipsilateral increase EMG profile during right side flexion for either ALBP ( $z = 0.5 - 1.5$ ,  $p = 0.134 - 0.6171$ ) or CLBP ( $z = 0 - 1.2$ ,  $p = 0.248 - 1$ ) group (table 5.1.11).

Table 5.1.11. Mean (sd) values of the muscle onset times (ms) for the ipsilateral increase EMG profile group for the movement of right side flexion (a negative value shows the muscle active following the onset of lumbar movement).

ALBP	Pre (ms)	Post (ms)
LLM	-45.7 (12.2)	-91.2 (46.2)
RLM	5.6(50.1)	-27.3 (111.7)
LIC	-34.1 (24.7)	-45.0 (17.0)
RIC	-9.3 (25.7)	-68.9 (103.3)

CLBP	Pre (ms)	Post (ms)
LLM	-65.6 (14.1)	-21.8 (34.2)
RLM	-54.3 (84.6)	29.4 (36.7)
LIC	-70.4 (31.8)	-20.9 (21.7)
RIC	-66.9 (49.6)	-49.8 (98.5)

Peak EMG activity was not significantly affected by pain relief in those who displayed the ipsilateral increased EMG profile during right side flexion for the ALBP ( $z = 0 - 0.5$ ,  $p = 0.6171 - 1$ ) or CLBP ( $z=0$ ,  $p = 1$ ) group (figure 5.1.11).

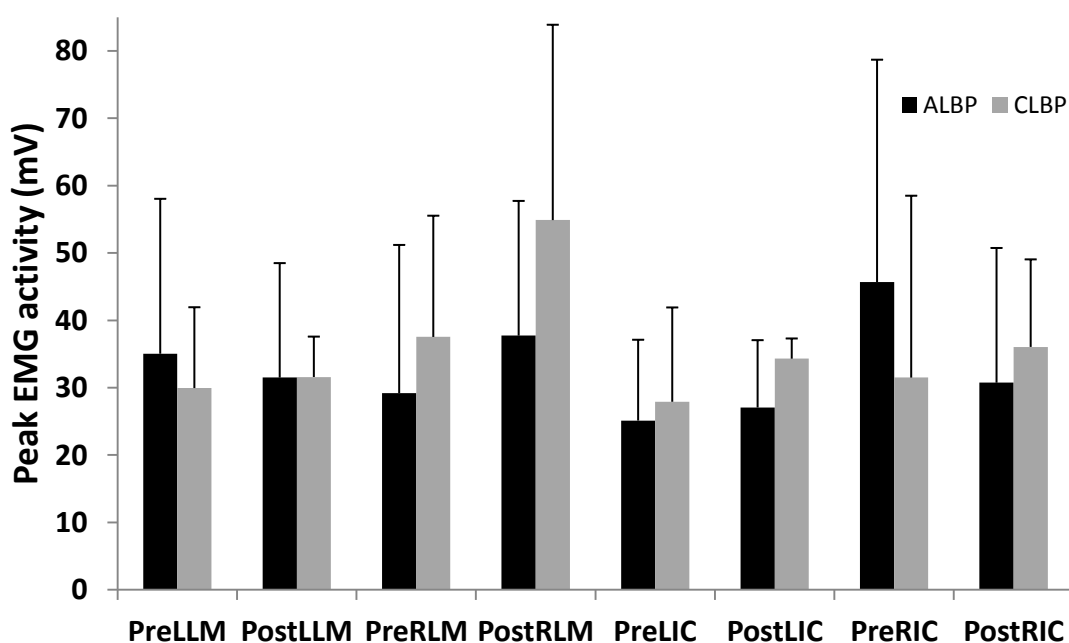
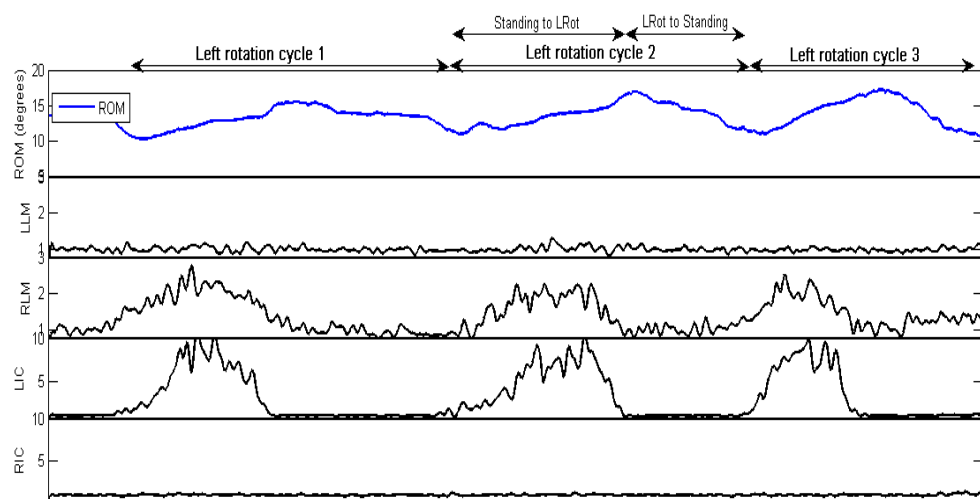


Figure 5.1.11. Mean pre and post pain relief peak EMG values for the ipsilateral increase EMG profile during right side flexion.

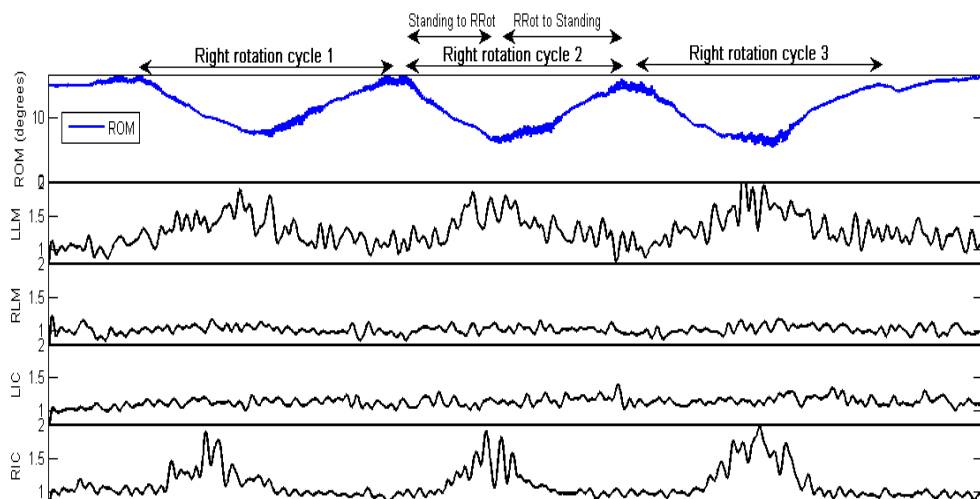
#### 5.1.4.5 Rotation

##### 5.1.4.5.1 Pattern

The EMG pattern observed during rotation was varied and therefore conclusions are limited. There was a tendency towards the iliocostalis working concentrically to pull the trunk around into rotation whilst activity is maintained in the opposite lumbar multifidus. This multifidus activity suggests a stabilising role to counteract the pull from iliocostalis (figure 5.1.12 (a) left rotation and (b) right rotation).



(a) Left rotation



(b) Right rotation

Figure 5.1.12. EMG profile during left (a) and right (b) rotation. LLM, left lumbar multifidus; RLM, right lumbar multifidus; LIC, left iliocostalis; RIC, right iliocostalis.

#### 5.1.4.5.2 Pain relief

All ALBP sufferers who reported pain relief did not alter the EMG profile for rotation.

CLBP sufferers displayed no alteration in the pattern of EMG in response to pain relief, with the exception of one individual who shifted from no evident pattern in RLM to a profile where RLM was active during right rotation to pull the trunk around into rotation (ipsilateral increase).

#### 5.1.4.6 Lifting

##### 5.1.4.6.1 Pattern

The EMG profile for lifting was commonly represented as similar to those presented for flexion, where different subjects appeared to demonstrate different EMG profiles, which generally fall into three types – (1) the increase profile (ALBP n = 7, CLBP n = 9) (figure 5.1.13), (2) the decrease profile (ALBP n = 3, CLBP n = 0) or (3) no pattern profile (ALBP = 2, CLBP n = 0). The increase EMG profile begins with the first muscle spike representing an eccentric period of muscle activity used to control the lowering of the trunk, followed by a period of low activity at end range flexion. Here the individual grasps the box and proceeds with a spike in EMG where the muscles work concentrically to pull the trunk and box up against gravity. A small amount of activity is retained in upright where the individual is holding the box in standing. Despite the additional weight of the box the lowering peak is similar in magnitude to the activity seen in lowering only the trunk. The box is released where a small spike in EMG profile is seen and the motion is complete with the concentric spike of the muscles to bring the trunk back to upright.

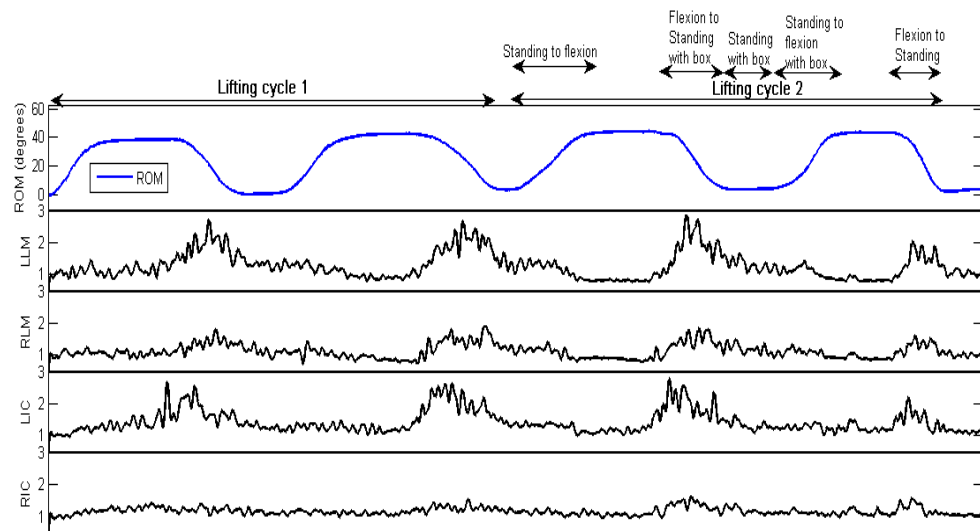


Figure 5.1.13. EMG profile during lifting – increase EMG profile, with kinematics at the top and EMG profiles below. LLM, left lumbar multifidus; RLM, right lumbar multifidus; LIC, left iliocostalis; RIC, right iliocostalis.

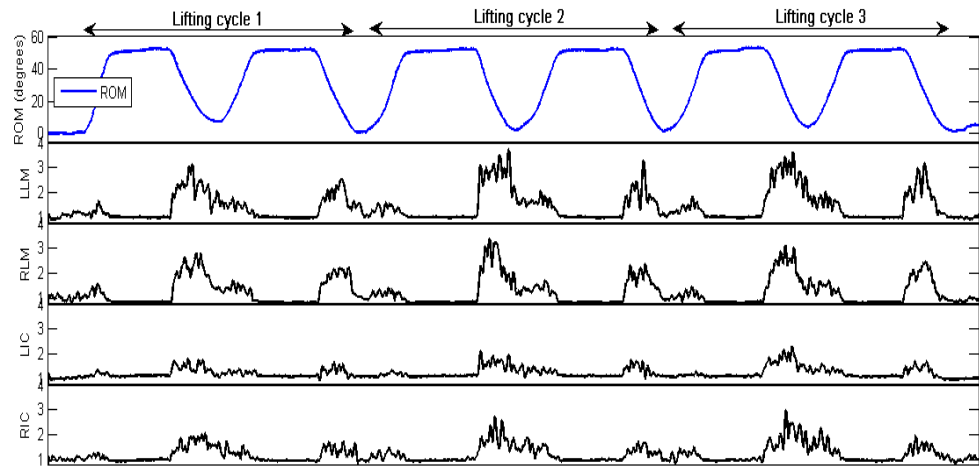
The cross-correlation coefficients were high for the increase EMG activity profile across both groups for the movement of lifting (table 5.1.12).

Table 5.1.12. Mean (sd) cross correlation between EMG profile and lumbar movement during lifting.

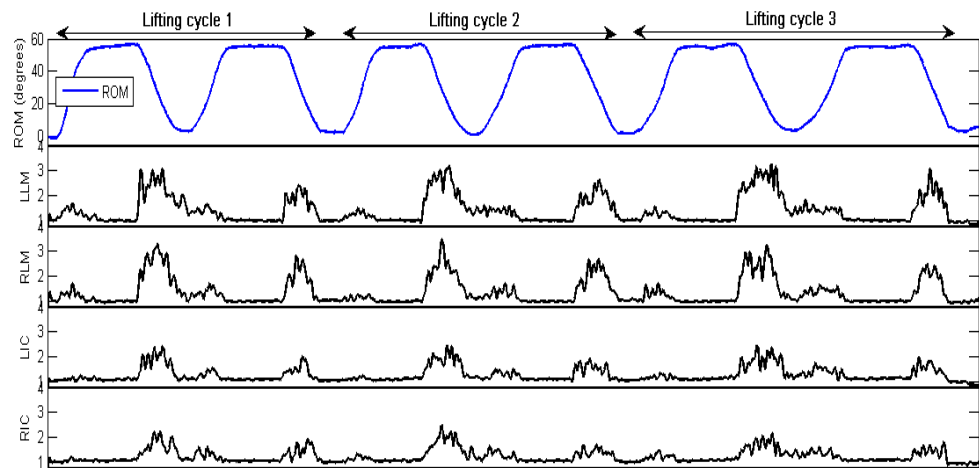
	ALBP	ALBP	CLBP
Lifting	Increase EMG profile	Decrease EMG profile	Increase EMG profile
n (%)	7 (58%)	3 (25%)	9 (100%)
Cross correlation			
LLM	0.78 (0.12)	0.78 (0.02)	0.81 (0.10)
RLM	0.77 (0.12)	0.77 (0.03)	0.79 (0.10)
LIC	0.77 (0.13)	0.77 (0.02)	0.81 (0.07)
RIC	0.77 (0.13)	0.77 (0.02)	0.81 (0.07)

#### 5.1.4.6.2 Pain Relief

ALBP and CLBP sufferers did not alter their EMG profile in response to pain relief (figure 5.1.14).



(a) Pre-pain relief lifting trials, VAS = 79



(b) Post-pain relief lifting trials, VAS = 64

Figure 5.1.14. Effect of pain relief on the lifting EMG profile – no change is demonstrated. (a) Participant lifting trials prior to (VAS = 79) and following (b) pain relief (VAS = 64).

EMG onset was not significantly affected by pain relief in those within the increase EMG profile during lifting for either ALBP ( $z = 0 - 0.76$ ,  $p = 0.459 - 1$ ) or CLBP ( $z = 0 - 0.67$ ,  $p = 0.505 - 1$ )

group with the exception of LLM which showed a significantly later onset ( $z = 2$ ,  $p = 0.045$ ) (table 5.1.13). Furthermore pain relief had no effect on the muscle onset time for the decrease EMG profile ( $z = 0$ ,  $p = 1$ ).

Table 5.1.13. Mean (sd) values of the muscle onset times for the increase EMG profile group for lumbar movement during lifting (a negative value shows the muscle active following the onset of lumbar movement).

	Increase EMG profile		Decrease EMG profile	
ALBP	Pre (ms)	Post (ms)	Pre (ms)	Post (ms)
LLM	-23.5 (21.4)	-23.4 (16.2)	4.9 (35.6)	10.0 (52.0)
RLM	-38.5 (29.8)	-17.4 (24.9)	10.0 (23.8)	31.0 (25.7)
LIC	-18.3 (13.6)	-17.4 (9.6)	4.4 (30.8)	15.5 (50.8)
RIC	-13.6 (24.5)	-11.1 (22.3)	5.3 (30.7)	8.0 (44.1)
CLBP	Pre (ms)	Post (ms)		
LLM	-9.8 (23.7)	-17.1 (26.2)*	-	-
RLM	-6.4 (30.9)	-22.0 (19.1)	-	-
LIC	-14.2 (15.6)	-17.6 (14.4)	-	-
RIC	-12.4 (20.0)	-14.0 (28.6)	-	-

$p < 0.05$ .

Peak EMG activity was not significantly affected by pain relief in those who displayed the increased EMG profile during lifting for the ALBP ( $z = 0 - 0.76$ ,  $p = 0.450 - 1$ ) or CLBP ( $z = 0 - 1.33$ ,  $p = 0.182 - 1$ ) group (figure 5.1.15). Furthermore peak EMG activity was not affected by pain relief in the decrease EMG profile group ( $z = 0 - 1.15$ ,  $p = 0.248 - 1$ ).



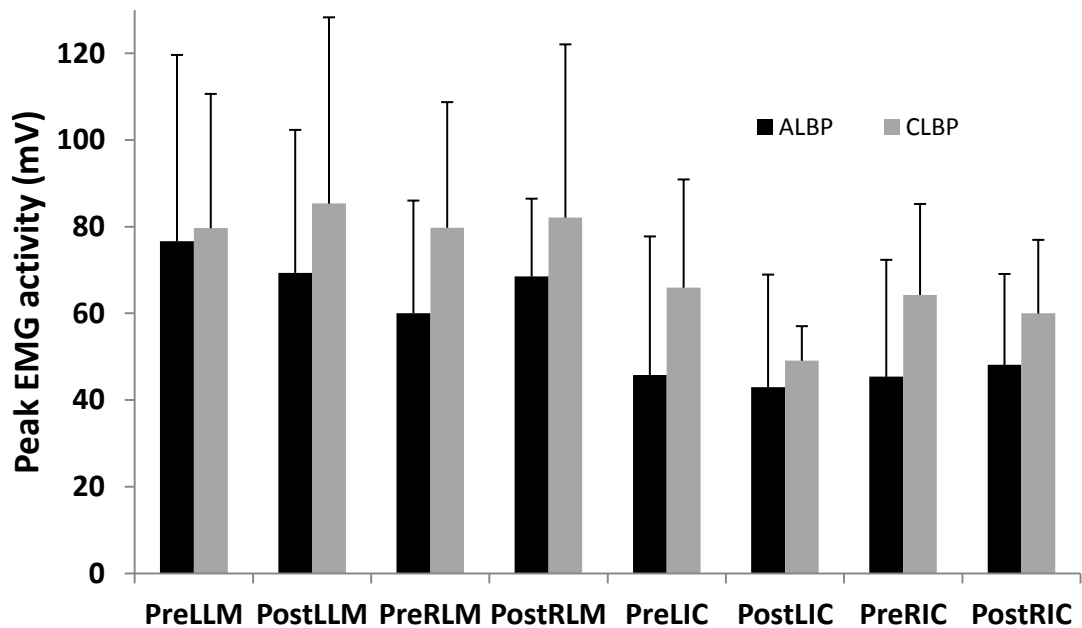


Figure 5.1.15. Mean pre and post pain relief peak EMG values for the increase EMG profile during lifting.

#### 5.1.5 Discussion

The present study examines the effects of pain relief on EMG during spinal movements in acute and chronic LBP sufferers. The measurement technique was found to result in reliable data, demonstrating good similarity for repeated EMG envelopes as well as highly reliable peak magnitude values. The results show in general that specific EMG profiles cross correlate well with movements of the lumbar spine and pain relief did not alter onset, peak magnitude or pattern of EMG activities.

Cross correlation analysis enables the determination of the strength of correlation between the EMG linear envelope or profile and the lumbar movements. Correlations were investigated for those EMG profiles demonstrating activity change during motion as the no pattern profiles demonstrate no peaks and troughs and therefore would be difficult to correlate with spinal movement. Moderate to good correlations were evident for all movements with the weakest

correlation being evident for extension. These slight differences in correlation are likely to be due to subtle changes in muscle activation patterns and kinematic profiles (Shum et al., 2005a).

The EMG profile patterns suggest a link between kinematic variables and EMG activity profiles, where higher temporal values were observed to coincide with specific EMG profiles. It is likely that increases in activity are required to increase the velocity and accelerations of the trunk during spinal motions. The absence of a flexion relaxation response is something commonly associated with LBP sufferers and was identified in some of the individuals in this study, but not all. It was most commonly seen in ALBP sufferers and was associated with lower kinematic variables. It is therefore likely that the absence of such a relaxation response identified in other studies may reflect deficits in lumbar kinematics (Geisser et al., 2005). Similar findings were evident for the movement of side bending where greater velocities and accelerations, but not ROM, were associated with particular EMG profiles. It is not clear whether clinical targets should focus on muscle training to alter the EMG profiles or whether restoration of lumbar kinematics would result in the automatic attenuation of the EMG profiles, however the results of this study do suggest that pain relief does not result in an automatic alteration of the EMG profile.

This study assessed the effects of pain relief on EMG onsets relating to spinal movements. It is important to note that complete pain relief was not achieved, however partial pain relief did not result in change in lumbar multifidus and iliocostalis onset timing. It is important to note that EMG profiles which demonstrated no pattern were not included as correlation to movement would be poor and therefore determination of onset would be difficult. Previous studies investigating the effects of pain on muscle onsets have utilised experimentally induced pain models, rather than models of pain relief as used in this study. These studies have demonstrated no significant difference in superficial lumbar multifidus or erector spinae onset

in response to experimentally induced pain (Moseley et al., 2004). These studies investigated muscle onsets in response to rapid arm movement, however the results are mirrored by those of the current study during functional movements. Zedka et al., (1999) also demonstrated reflex activation was unaffected by induced pain. Previous studies have shown differences in onset relating to sudden loading between individuals with CLBP and controls, however these individuals had sciatic symptoms which represents a different sub-population of LBP sufferers (Leinonen et al., 2001; Shum, Crosbie, & Lee, 2007b).

The multifidus muscle has been a suggested target for therapeutic interventions associated with LBP despite the evidence of its involvement not being clear (MacDonald, Moseley, & Hodges, 2006). The results from this study suggest that clinicians should not expect an immediate alteration in muscle onset time in response to partial pain relief.

This study also investigated the effects of partial pain relief on peak EMG amplitude of paraspinal muscle activity and demonstrates that peak EMG amplitudes were not influenced by the pain relief. These findings are consistent with previous studies which have shown no change in muscle performance during isokinetic dynamometry following pain relief (Holm et al., 2000). These findings therefore suggest that partial pain relief does not lead to an immediate alteration in peak muscle amplitudes. However this is in contrast to others who have shown an increase in muscle performance (Sorenson test) following pain relief (Rashiq et al., 2003). Unfortunately neither of these studies incorporated EMG measurements and the relationship between EMG activity and either isokinetic dynamometry or the Sorenson test is not well understood making it difficult to predict how these changes would affect peak EMG amplitudes. Furthermore the Sorenson test is a test of muscle endurance therefore limiting the comparability of the results to the present study.

This study was limited to spinal movements and therefore it may not be possible to extrapolate the results to other functional tasks. The sample size was small and limited to those suffering from pain confined to the low back, therefore the findings may not be generalisable to other LBP sub-populations. Further limitations are that this study investigated the immediate effects of pain relief only. This was done to minimise the influence of other factors which may affect the muscle activity in an attempt to isolate the investigation to the effects of pain relief alone, however the findings may not be reflective of prolonged pain relief.

#### **5.1.6 Conclusion**

This study showed that the EMG activities from the lumbar region, including lumbar multifidus and iliocostalis, are reliable and specific EMG profiles are identifiable. These EMG profiles cross-correlate well to lumbar kinematics. Cross-correlation lags demonstrated onsets were unaffected by partial pain-relief. Partial pain-relief also did not alter peak EMG amplitudes or EMG profiles during spinal movements. Therefore clinicians wanting to influence the muscles of multifidus and iliocostalis should not expect an immediate automatic alteration of temporal-spatial EMG parameters following partial pain relief.

## 5.2 Summary of results and key findings

The results of this chapter demonstrate that EMG linear envelopes produced during lumbar movements offer good repeated measures reliability as shown by good CMC and ICC values.

This chapter presents the use of EMG profile patterns as a way to describe the muscle functions relative to kinematic behaviour. This method enables the clinician to quickly understand the relationship between muscle function and trunk kinematics. The identified patterns commonly represent either increases or decreases in muscle activity and these definable patterns were often associated with greater kinematic values. Conversely the EMG profiles which display no pattern and do not deviate greatly from a consistent level of activity were often related to lower kinematics values.

The results demonstrated that partial pain relief had no effect on muscle onset times or peak amplitudes. This suggests that partial pain relief does not result in an automatic alteration of lumbar muscle function for the muscles of multifidus and iliocostalis. This finding means that clinicians should not expect an alteration in lumbar muscle function following interventions achieving partial pain relief.

### Summary of key findings

- The reliability of repeated measures EMG during spinal movements is good for both profile and peak values.
- EMG behaviour can be classified based on the shape of the EMG profile.
- Definable EMG profiles may be related to greater lumbar kinematics values.
- Partial pain relief did not alter the timing of muscle onset or the peak EMG magnitude.

# **Chapter 6:**

## **General discussion**

## 6.1 General discussion

This thesis set out with the aim of determining whether pain is the cause of altered movement and muscle functions in back pain sufferers? This was achieved by employing an effective method of simple pain relief before and after the detailed analysis of motion and motor patterns in sufferers of movement evoked LBP.

The assessment and management of LBP often involves the observation of motion and motor patterns which are used as a framework for therapeutic intervention. Therefore the ability to objectively measure motion within the clinical environment is an important aim for clinicians. It was determined from a review of the literature that currently available methods of curvature measurement have significant limitations regarding the detail of kinematic information available. Referring back to table 2.1, it is evident that commonly this limitation was in the inability to measure dynamic kinematics and curvature resulting in movement behaviour through time information being unavailable. Clinician's current employ assessment methods either based on visualisation of motion or static end range measurements. Daily functional tasks rarely require the spine to move through full range of motion and range of motion values are not well correlated to functional ability or disability (Parks, Crichton, Goldford, & McGill, 2003). Therefore range of motion alone is of limited help to the clinician. In light of this clinical models have focussed on alternate kinematic variables such as pattern, coordination and sequencing of movement. These variables are extremely difficult to observe by eye and therefore clinic based motion analysis technologies are required to give clinician a more detailed objective picture of movement behaviour. Furthermore higher order kinematics such as velocity and acceleration, correlate with loss of function and disability (Marras, Davis, Ferguson, Lucas, & Gupta, 2001; Marras et al., 1999; Marras, Lewis, Ferguson, & Parnianpour, 2000b; Marras et al., 1995; Marras & Wongsam, 1986) and are slower to recover in LBP populations (Ferguson, Marras, & Gupta, 2000). In order for clinicians to be truly able to observe detailed kinematics, instrumentation of the individual with LBP is required. Moreover

objective determination of treatment effect is important in the process of restoring normal spinal functions. Therefore there is a need for detailed motion analysis within the clinical environment to better inform the clinician of the spinal movement behaviour for the use in assessment and treatment.

The initial phases of this research involved the development of using new technology for the clinical evaluation of lumbar curvature and kinematics. The results of this thesis have demonstrated for the first time that fibre-optics may be a viable option to dynamically measure lumbar curvature. The fibre-optic device can overcome the limitations by providing dynamic curvature measurement. This dynamic measurement was shown to be possible for different regions of the lumbar spine and therefore could provide a regional breakdown of curvature. This could be of great use to the clinician attempting to determine the curvature at different regions of interest. Examples may include following surgery where the curvature at a specific region could be measured through a range of functional tasks or where a region is known to be affected by pathology, such as spondylolithesis or scoliosis, and the interest is in how this affects the spinal curvature during movement. The fibre-optic system does not have environmental limitations commonly associated with other systems (Milne, Chess, Johnson, & King, 1996; Ng, Burnett, Campbell, & O'Sullivan, 2009), making it an excellent option for regional dynamic curvature measurement.

The results also demonstrate that, as the system can resolve dynamic curvature measurement, it can also be used to measure sequencing of lumbar curvature change. Although this is not unique to this device, the advantage would be that sequencing of curvature change could be determined for any region of interest along the spine. Sequencing does not appear frequently within the literature, however despite this models of clinical management often involve the identification and modification of sequencing of motion related to lumbar movement, to which clinical success has been attributed (Dankaerts, O'Sullivan, Burnett, & Straker, 2007; O'Sullivan,



2000). In light of this clinical trend, there is a need provide objective clinical measurement of such kinematics.

Current laboratory based measurement methods used to assess lumbar kinematics have known environmental constraints (Milne et al., 1996; Ng et al., 2009) making them less than ideal for clinical based motion capture. Furthermore, reporting of lumbar kinematics is often confined to ROM, however as reported previously, higher order kinematics of velocity and acceleration are known to be of great importance in LBP (Marras, Allread, Burr, & Fathallah, 2000a; Marras et al., 1993). The findings of this thesis illustrate that inertial sensors can be used to assess lumbar kinematics. This work extends the current knowledge as this is the first time such technology has been used within a clinical population and within a clinical environment. This technique provides clinicians with a method of analysing three-dimensional kinematics of the lumbar spine. The findings further extend the current knowledge by the inclusion of reliability extended to the higher order kinematics of velocity and acceleration. This is clinically important due to these variables being related to LBP risk and LBP reporting (Marras et al., 2000a; Marras et al., 1993) as well as these variables offering better discriminatory ability than just ROM alone (Marras et al., 1986). It has also been shown that these variables are slower to recover during the resolution of LBP making them a sensitive measure of functional recovery (Ferguson et al., 2000). The sensors are fully portable, not constrained by the environment and now that reliability in clinical populations has been shown their use is open to a wide variety of possible applications. Examples of such may include the spinal kinematic profiling of particular occupational or sporting tasks where the motion characteristics including velocities and accelerations can be investigated. This information could then be applied to the individual recovering from LBP where higher order kinematics are used to monitor the recovery of spinal function, in order to determine if the functional capacity required for particular occupational or sporting tasks has been achieved.

As the findings have demonstrated the ability to yield velocity measurements this work has demonstrated the application of spatial domain analysis. This approach has been utilised previously to measure the coordination between moving segments, expressed as phase angle (Shum, Crosbie, & Lee, 2005a), however the current work extends this by proposing a new method of spatial domain analysis where motion irregularity is quantified. This additional analysis is achieved without further data collection and enables the clinician to explore the movement trajectory as a measure of movement coordination. This analysis technique was demonstrated to be reliable and can be used to highlight phases of the motion associated with greater irregularity. Examples of an application may include the use of this trajectory plot to provide insights into movement irregularity associated with a functional or sporting task. This phase may then be targeted to either increase or decrease the irregularity depending on the desire of the clinician/coach.

A key clinical advantage of this motion analysis option is the application of real-time feedback. Clinicians often attempt to alter movement strategies (O'Sullivan, 2000) and the use of biofeedback in real-time may prove very useful for this objective. This can be applied for any of the variables investigated and even extend to the spatial domain where feedback can be used to alter movement irregularity. This may prove very useful for aspects where rehabilitation may focus on using higher velocities and accelerations in an attempt to improve lingering functional deficits (Ferguson et al., 2000; McGill et al., 2003; Shum, Crosbie, & Lee, 2009).

These novel motion analysis methods are believed to offer a solution to the limitations outlined in currently available clinical motion analysis methods. It is possible that such technology will become commonplace within the clinical environment providing additional information not currently available to assist in the assessment and management of LBP.

The observation of movement alterations has been commonly documented in LBP sufferers (Marras et al., 2001; Marras et al., 1999; Marras et al., 1986; Novy, Simmonds, Olson, Lee, & Jones, 1999; Shum et al., 2005a; Shum, Crosbie, & Lee, 2005b; , 2007; Wong & Lee, 2004), and

the results of this study demonstrate that there may be differences between the movement profiles of ALBP and CLBP sufferers. ALBP sufferers demonstrate less peak curvature during flexion and lifting along with a trend towards less curvature during extension. Similar trends were evident for peak kinematic variables during sagittal plane motions (flexion, extension and lifting). It is logical to assume that greater curvature formation would result in greater range of motion, however it also appears that these greater curvatures and ROM are observed along with greater higher order kinematics. These findings therefore suggest that CLBP sufferers have a tendency to slightly outperform ALBP during sagittal motions. These findings continue with similarity to the EMG findings. The prevalence of an EMG pattern consistent with an increase EMG profile during flexion was slightly more common in the CLBP group (n=8) compared to the ALBP group (n=5). This pattern relates to the display of greater ROM, negative velocity and acceleration by the CLBP group, which suggests that the return from the flexed position is associated with greater EMG peaks and may be an important feature of the increase EMG profile for flexion. Furthermore the increase EMG profile was observed in all CLBP sufferers (n=9) during lifting, whereas a profile other than the increase EMG profile was produced in five ALBP suggesting that again the higher performing group (CLBP) displaying greater kinematics was linked to the creation of EMG profiles with clearly defined activation peaks.

However just measuring movement objectively is insufficient in truly understanding why LBP sufferers adopt the movement patterns they do. Management strategies often involve either targeting pain relief to alter spinal biomechanical functions (Jette, Smith, Haley, & Davis, 1994) or directly aim to influence movements and motor behaviour (O'Sullivan, 2005) with no clear rationale as to which method should be used. Therefore if the clinical aim of treatment is to restore the biomechanical behaviour of the spine then it is imperative to identify the underlying cause or mechanism responsible for these alterations so that clinical management can begin to be rationalised.

This thesis set out with the aim of determining whether pain relief is associated with changes in movement and muscle functions in back pain sufferers. It should be acknowledged that the simple pain relief method employed in this research only resulted in partial pain relief and that this may therefore not reflect the situation associated with complete abolition of pain. The results demonstrate that partial pain relief resulted in a decrease in peak lumbar curvature for ALBP sufferers during flexion and lifting for the whole lumbar spine and for lifting in the lower lumbar spine. It had no effect on extension or the CLBP sufferers' curvature measurements. Partial pain relief had no effect on lumbar kinematics including range of motion, angular velocity and angular acceleration and did not, on the whole, alter EMG. This suggests that either the amount of pain relief was insufficient or that a mechanism other than pain may be responsible for the maintenance of movement alteration in LBP sufferers.

Despite this overall effect some individuals did respond and there were some additional interactions worthy of note between the variables when investigating these individual responders. The definition of responders for this thesis was presented in chapters 3 and 4 but is re-presented here (table 6.1).

Table 6.1. Gains required to be defined as a responder

	Curvature	Kinematics
Flexion/Lifting	$>3^{\circ}$	$>3^{\circ}, 4^{\circ}\text{s}^{-1}, 8^{\circ}\text{s}^{-2}$
Extension/Side-Flexion	$>2^{\circ}$	$>2^{\circ}, 3^{\circ}\text{s}^{-1}, 6^{\circ}\text{s}^{-2}$
Rotation	na	$>1^{\circ}, 2^{\circ}\text{s}^{-1}, 4^{\circ}\text{s}^{-2}$

Table 6.2 displays the change in curvatures and kinematics for each group.

Table 6.2. Change in variables following pain relief.

ALBP	Peak curve (°)	ROM (°)	PosVel (°s <sup>-1</sup> )	NegVel (°s <sup>-1</sup> )	PosAcc (°s <sup>-2</sup> )	NegAcc (°s <sup>-2</sup> )
Flex	-3.1 (2.1)	0.1 (4.3)	0.3 (5.2)	4.8 (4.7)	6.1 (12.3)	6.9 (11.7)
Lift	-2.7 (2.6)	0.2 (5.5)	1.5 (5.7)	0.3 (5.2)	3.0 (10.9)	0.0 (13.0)
Ext	5.2 (9.9)	-0.3 (2.2)	0.3 (4.4)	0.6 (1.9)	1.7 (9.6)	1.8 (9.2)
CLBP	Peak curve (°)	ROM (°)	PosVel (°s <sup>-1</sup> )	NegVel (°s <sup>-1</sup> )	PosAcc (°s <sup>-2</sup> )	NegAcc (°s <sup>-2</sup> )
Flex	0.4 (4.8)	0.1 (6.6)	2.6 (4.4)	1.1 (6.1)	7.1 (8.1)	0.5 (7.9)
Lift	1.6 (3.6)	0.4 (4.5)	-0.2 (9.7)	-0.6 (4.8)	3.2 (16.3)	3.3 (12.6)
Ext	-0.8 (6.0)	1.0 (2.8)	1.8 (4.0)	2.1 (4.8)	4.8 (8.4)	5.1 (8.0)

Flex, Flexion; Ext, Extension; ROM, Range of motion; PosVel, positive velocity; NegVel, negative velocity; PosAcc, positive acceleration; NegAcc, negative acceleration

It is interesting that the changes in curvature were a reduction for flexion and lifting which is not seen in the kinematic variable of range of motion. This suggests that a reduction in peak curvature was achieved following pain relief in the ALBP group however the same group did not demonstrate any change in lumbar displacement. Furthermore little relationship could be established between change in curvature and kinematics for responders. Only one individual (ALBP) was defined as a responder for both curvature and all kinematic variables and this was for the movement of lifting. One individual in the CLBP group demonstrated a response for curvature and three of the kinematic variables and this was for the movement of flexion. Despite this no other individuals, defined as responders for curvature change following pain relief, could be defined as having 'responded' regarding their kinematics values. It is logical to assume that an alteration in curvature would result in an alteration in range of motion also, however this was not the case.

These findings further suggest that pain relief induced changes in curvature do not result in changes in higher order kinematics suggesting that these variables are distinctly different. The clinical significance of such is that interventions resulting in curvature changes may not also automatically result in changes in movement velocity and acceleration.

The individuals demonstrating a definable EMG profile often display greater kinematic values suggesting a link between these two variables. It is likely that greater muscle activities are required to move the trunk at greater velocities and accelerations. Clinically it may be possible that an alteration in one of these variables results in an automatic alteration of the other. An example would be the restoration of the increase EMG profile seen during flexion by increasing the kinematic variables of ROM, angular velocity and angular acceleration. However this idea is not supported by the results from this thesis. In the individuals who altered their EMG pattern from no pattern to a definable pattern, very few of the kinematic and curvature variables altered by a clinically sufficient amount. This suggests there is no immediate link between the alteration of lumbar kinematics and curvature with the alteration of EMG profile. These findings may suggest that in those individuals who did alter their EMG profile did so in such a way as to not affect the peak kinematic variables. An example of such maybe the development of co-contraction in the muscles being measured but also in other muscles (such as the abdominals) which may cause an increase in activation profile but may not increase the peak kinematics or curvature.

The results of this thesis provide data which does not support two commonly reported theories regarding the relationship between movement and motor functions and pain. Firstly it is believed that altered movement profiles serve as a protective function. This proposed adaptive compensation is believed to afford the sufferer with an alternate movement strategy that is less provocative of pain (Shum et al., 2005a; 2005b; 2007). If this were the case then logically it would be assumed that the abolition of pain would result in a change in kinematic

profile. The current study did achieve a significant reduction in pain however it did not abolish the pain entirely. Therefore it is not known whether the abolition of pain would result in similar findings to the current study achieving only partial pain relief. This is the first time however that such a study has been completed where simple pain relief, through the use of pharmacological agents, was employed and functional movements investigated. It is suggested that the movement strategies may be associated with some unknown mechanisms or mechanisms which could have been altered by pain, but pain did not influence kinematics directly. Previous work has demonstrated gains in movement velocity in response to heat induced pain relief (Simmonds & Rebelo, 2003), which unfortunately only confuses the issue as heat may alter pain as well as tissue properties. This study focused on the immediate effects only and therefore it is unclear as to whether more time and specifically more 'rehearsal' of movements with less (or no pain) would result in modification of the movement strategy through a processes of gradual adaptation. In any case, the present results demonstrate that pain relief did not immediately increase lumbar curvature, kinematics or muscle function in back pain sufferers as would be predicted by the adaptive compensation theory.

The second belief amongst clinicians is that therapeutic techniques targeted at pain relief (eg Grade I/II spinal mobilisation) can bring about alterations in curvature, kinematics and muscle functions (Jette et al., 1994). This thesis helps rationalise our body of knowledge by disputing this belief which may not be necessarily true. Indeed, therapeutic techniques which target pain relief will not, based on these results, alter these biomechanical functions of the lumbar spine. Therefore if the therapeutic aim is to restore normal biomechanical function of the lumbar spine then simple pain relief will not serve the purpose. This raises important questions as to the mechanisms behind therapeutic interventions with reported immediate effects on lumbar kinematics, such as manipulation (Childs, Piva, & Erhard, 2004; Lehman & McGill, 2001). The mechanisms underlying these immediate effects are likely to be due to biomechanical and/or physiological changes which are brought about by the treatment, although such treatments may lead to pain relief too.

## **Chapter 7:**

# **Conclusions and recommendations**



## **7.1 Conclusion**

This thesis has demonstrated that dynamic curvature of the lumbar spine can be measured using fibre-optic sensors. The reliability and validity was good to excellent as compared to a video based system and may therefore offer a method of dynamic curvature measurement within the field of biomechanics.

The results demonstrate that lumbar curvature behaviour of small spinal regions can be investigated. This is achieved without additional data capture and can include the sequencing of curvature change, a variable often discussed in clinical management models.

The results illustrate that inertial sensors are reliable for the measurement of lumbar kinematics in a clinical environment. Furthermore this reliability has now been demonstrated for the higher order kinematics of velocity and acceleration.

This thesis has offered a measurement method of movement irregularity through analysis in the spatial domain. The method was shown to be reliable and capable of determining movement irregularity for different segments of motion.

The results show that ALBP sufferers display less peak curvature than CLBP for flexion and lifting as measured by the fibre-optic device. ALBP sufferers display similar kinematics to CLBP sufferers, as measured by the inertial sensor system.

Partial pain-relief, as achieved in this study, did not attenuate ROM, angular velocity, angular acceleration, movement irregularity or EMG profile, muscle onset time and peak EMG activity

in ALBP or CLBP. Furthermore, neither acute nor chronic LBP sufferers were more likely to respond to pain relief.

## **7.2 Recommendations for future work**

- Extend dynamic curvature analysis to known spinal disorders such as scoliosis, spondylolithesis and osteoporosis and relate dynamic postural changes to functional activities.
- Investigate the discriminatory ability of spatial domain analysis.
- Investigate the importance of real time feedback using fibre-optic sensors or inertial sensors in clinical rehabilitation.
- Investigate the effects of altering sequencing of curvature change on evoked pain.
- Investigate the effect of repeated movements, which are likely to either increase or decrease the evoked pain and monitor the effect of this gradual change in pain on change in kinematics and muscle functions.
- Investigate the effect of pain relief over time on the kinematics and muscle activities.
- Determine the characteristics which are associated with those individuals who are responders to pain-relief to facilitate clinical decision making.
- Investigate other potential mechanisms which could be responsible for movement alterations in LBP.

### **7.3 Final conclusion**

This thesis has rationalised our current body of knowledge by providing new insights into the complex relationship between pain and movement in sufferers of LBP. The findings provide evidence to suggest that simple partial pain relief has no impact on lumbar biomechanics challenging some popularly held beliefs. Further research work should aim to identify other potential mechanisms responsible for movement alterations in LBP sufferers. It is hoped that the improvement in our understanding of back pain, as a result of the work of this thesis, would help reduce the socioeconomic impact of back pain and contribute to further development in the clinical management of many back pain sufferers.

## **Appendix 1.0 Checklist for measuring study quality (Downs and Black 1998)**

1. Is the hypothesis/aim/objective of the study clearly described?
2. Are the main outcomes to be measured clearly described in the Introduction or Methods section?
3. Are the characteristics of the patients included in the study clearly described?
4. Are the interventions of interest clearly described?
5. Are the distributions of principal confounders in each group of subjects to be compared clearly described?
6. Are the main findings of the study clearly described?
7. Does the study provide estimates of the random variability in the data for the main outcomes?
8. Have all important adverse events that may be a consequence of the intervention been reported?
9. Have the characteristics of patients lost to follow-up been described?
10. Have actual probability values been reported (e.g., 0.035 rather than  $<0.05$ ) for the main outcomes except where the probability value is less than 0.001?
11. Were the subjects asked to participate in the study representative of the entire population from which they were recruited?
12. Were those subjects who were prepared to participate representative of the entire population from which they were recruited?
13. Were the staff, places, and facilities where the patients were treated, representative of the treatment the majority of patients receive?
14. Was an attempt made to blind study subjects to the intervention they have received?
15. Was an attempt made to blind those measuring the main outcomes of the intervention?
16. If any of the results of the study were based on “data dredging”, was this made clear?

17. In trials and cohort studies, do the analyses adjust for different lengths of follow-up of patients, or in case–control studies, is the time period between the intervention and outcome the same for cases and controls?
18. Were the statistical tests used to assess the main outcomes appropriate?
19. Was compliance with the intervention/s reliable?
20. Were the main outcome measures used accurate (valid and reliable)?
21. Were the patients in different intervention groups (trials and cohort studies) or were the cases and controls (case–control studies) recruited from the same population?
22. Were study subjects in different intervention groups (trials and cohort studies) or were the cases and controls (case–control studies) recruited over the same period of time?
23. Were study subjects randomised to intervention groups?
24. Was the randomised intervention assignment concealed from both patients and health care staff until recruitment was complete and irrevocable?
25. All non-randomised studies should be?
26. Was there adequate adjustment for confounding in the analyses from which the main findings were drawn?
27. Were losses of patients to follow-up taken into account?

## Appendix 1.1 Consent form



### PARTICIPANT CONSENT FORM

#### Title of Research Project:

Fibre-optics for spinal curvature measurement: Validity and Reliability

#### Brief Description of Research Project:

The aim of this study is to investigate the use of fibre-optic technology to measure movements and curvature of the low back. A fibre-optic ribbon will be attached to the back and movements measured which will then be compared to a video camera with reflective marker system.

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#### Consent Statement:

I agree to take part in this research, and am aware that I am free to withdraw at any point. I understand that the information I provide will be treated in confidence by the investigator and that my identity will be protected in the publication of any findings.

Name .....

Signature .....

Date .....

Please note: if you have a concern about any aspect of your participation or any other queries please raise this with the investigator. However if you would like to contact an independent party please contact the Dean of School (or if the researcher is a student you can also contact the Director of Studies.)

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**Consent Form**

**Project Aims:**

Investigate the effects of pain relief on movements and muscles of the back.

Study the differences between back pain sufferers with pain of recent onset and those which have more long standing pain.

- The researcher has explained to my satisfaction the purpose of the study and the possible risks involved.
- I have had the procedure explained to me and I have also read the information sheet. I understand the procedures fully.
- I am aware that I will be required to undertake movement tasks that may cause pain whilst having these movements monitored, then take oral analgesia and repeat these movement tasks.
- I understand that any confidential information will be seen only by the researchers and will not be revealed to anyone else.
- I understand that I am free to withdraw from the investigation at any time.
- I understand that if I agree to participate my GP will be informed.
- I agree to participant in this research project.

Name (please print)

.....

Signed

.....

Date

.....

**consent form 28-06-08 ver2**

### Appendix 1.3 Tampa Scale of Kinesiophobia

1. I'm afraid that I might injure myself if I exercise.

Strongly disagree      1      2      3      4      Strongly agree

2. If I were to try to overcome it, my pain would increase.

Strongly disagree      1      2      3      4      Strongly agree

3. My body is telling me I have something dangerously wrong.

Strongly disagree      1      2      3      4      Strongly agree

4. My pain would probably be relieved if I were to exercise.

Strongly disagree      1      2      3      4      Strongly agree

5. People aren't taking my medical condition seriously enough.

Strongly disagree      1      2      3      4      Strongly agree

6. My accident has put my body at risk for the rest of my life.

Strongly disagree      1      2      3      4      Strongly agree

7. Pain always means I have injured my body.

Strongly disagree      1      2      3      4      Strongly agree

8. Just because something aggravates my pain does not mean it is dangerous.

Strongly disagree      1      2      3      4      Strongly agree

9. I am afraid that I might injure myself accidentally.

Strongly disagree      1      2      3      4      Strongly agree

10. Simply being careful that I do not make any unnecessary movements is the safest thing I can do to prevent my pain from worsening.

Strongly disagree      1      2      3      4      Strongly agree

11. I wouldn't have this much pain if there weren't something potentially dangerous going on in my body.

Strongly disagree      1      2      3      4      Strongly agree

12. Although my condition is painful, I would be better off if I were physically active.

Strongly disagree      1      2      3      4      Strongly agree



13. Pain lets me know when to stop exercising so that I don't injure myself.

Strongly disagree      1      2      3      4      Strongly agree

14. It's really not safe for a person with a condition like mine to be physically active.

Strongly disagree      1      2      3      4      Strongly agree

15. I can't do all the things normal people do because it's too easy for me to get injured.

Strongly disagree      1      2      3      4      Strongly agree

16. Even though something is causing me a lot of pain, I don't think it's actually dangerous.

Strongly disagree      1      2      3      4      Strongly agree

17. No one should have to exercise when he/she is in pain.

Strongly disagree      1      2      3      4      Strongly agree

#### **Appendix 4.0: Fibre-optic (Shaptetape) attachment and processing trials**

As the fibre-optic system is a ribbon of sensors fully attached throughout its length, attachment to the human presents a challenge. The string of sensors have no elastic qualities therefore to simply adhere it to the skin would prevent movement. A series of attachment trials was conducted to investigate the optimal attachment method.

##### **Trial 1**

This utilised the back attachment kit provided by the company. This consisted of four individual elasticised straps attached circumferentially around the trunk spaced evenly apart. The region over the spine had a small piece of Velcro onto which a fabric tube was attached. The shaptetape passed into fabric tube allowing the shaptetape to slide through the tube to accommodate movement. The system worked well for lumbar flexion however on return the shaptetape failed to slide back in to the tube resulting in buckling. Buckling of the whole system was seen during lumbar extension motion. It was therefore concluded that this system failed to hold the shaptetape in place to follow the contour of the spine the system was discarded.

##### **Trial 2**

In light of the problem of re-entry of the shaptetape into the tube a modified system was trialled, where two pieces of elastic were bridged with small rollers, resembling a ladder. This was adhered to the skin using elastic bandage along each long edge and the base of the shaptetape was firmly fastened to the skin over the first sacral vertebrae (S1) (see fig. A4.1).

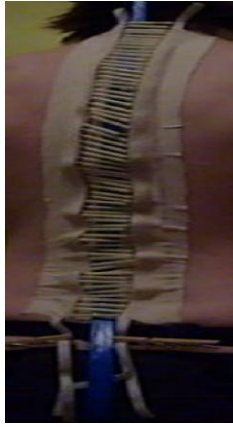


Figure A4.1. Trial 2 for fitting the fibre-optic device to the participant.

As the subject flexed the elastic allowed free motion and the shapetape followed the contour of the back. On return from flexion the shapetape passed back into the attachment system without problem. However this system failed to tightly constrain the shapetape to the back and was extremely difficult to attach with only one person. In light of these limitations it was concluded that this system was not optimal.

### Trial 3

In order to overcome the failure to conform closely to the contours of the back a large modified elastic tubular bandage was constructed. This was stitched to create a channel for the shapetape to pass through and was adjustable through the addition of Velcro straps sown into the front. A small hole was cut allowing the adhesion of the base sensor to the skin overlying S1 (see fig. A4.2).

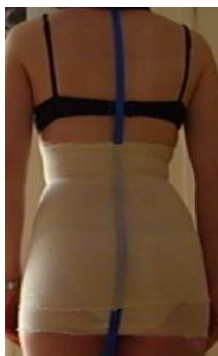


Figure A4.2. Trial 3 for fitting the fibre-optic device to the participant.

During motion this system enabled the shapetape to follow the contours very closely both with flexion and its return as well as extension and its return; however the limitation is the inability to allow any other analytical systems to be used simultaneously, such as emg.

#### Current system

In order to overcome the limitations of previous systems the following design characteristics were drawn up:

1. The base sensor needed to be directly adhered to the skin overlying the S1 spinous process.
2. The shapetape system needed to be able to slide over the skin of the spine to conform to the spinal shape.
3. The method of attachment must not interfere with spinal motion.
4. The attachment system should be elastic to prevent buckling of the shapetape on return from flexion and in extension.
5. The system should allow other technologies, such as emg to be attached if required.
6. Attachment should be quick and simple requiring one person only.

The system comprised of a tube constructed out of elastic tubular bandage. This was narrow enough to fit the width of the shapetape without excessive room for lateral sliding movement, with its elastic properties being evident in the vertical direction. The shapetape is passed into the tube up to the location of the first sensor, which is fastened to the skin with double sided tape (wig tape, Wigs and Pieces, UK) overlying the S1 vertebrae. The box positioned 120mm from the first sensor would be attached with velco to a pair of shorts. This would then support the weight of the box allowing enough slack to accommodate the flexion movement so that no pulling would be felt by the subject. The elasticated tube would be maintained flush to the skin with a length of elastic adhesive bandage (Tenoplast, PhysioMed, UK). Importantly these are applied with the subject in near full flexion and whilst the elastic tensioned. Therefore on

return to upright the bandage just returns to its original shape rather than buckling. During motion the shapetape slides through the tube which is able to cope with length changes required to complete the motions. As the system is attached under tension and in flexion it remains flush and following the contours of the skin with only a thin layer of material between the shapetape and the skin of the back. It leaves the rest of the skin on the back free for attachment of emg if required and can be applied without requiring assistance. Therefore this system meets the design brief and will be used throughout this thesis (figure A4.3).



Figure A4.3. Current attachment method for fitting the fibre-optic device to the participant.

## Curve fitting

As data were to be spline fitted it was necessary to determine which spline offered the best fit.. All raw data were analysed using Matlab (Mathworks, R2008b). Raw sagittal plane data for the shapetape system was sequentially fitted with a variety of functions including polynomials and splines to determine the most appropriate curve fitting tool. A cubic spline and cubic hermite spline visually offered the best fit with no residuals and further exploration determined the piecewise cubic hermite interpolating polynomial offered the best fit. The reason behind this can be visualised in figure A4.6, where the shape of the lower lumbar spine is incorrectly represented by the cubic spline in the Vicon model.

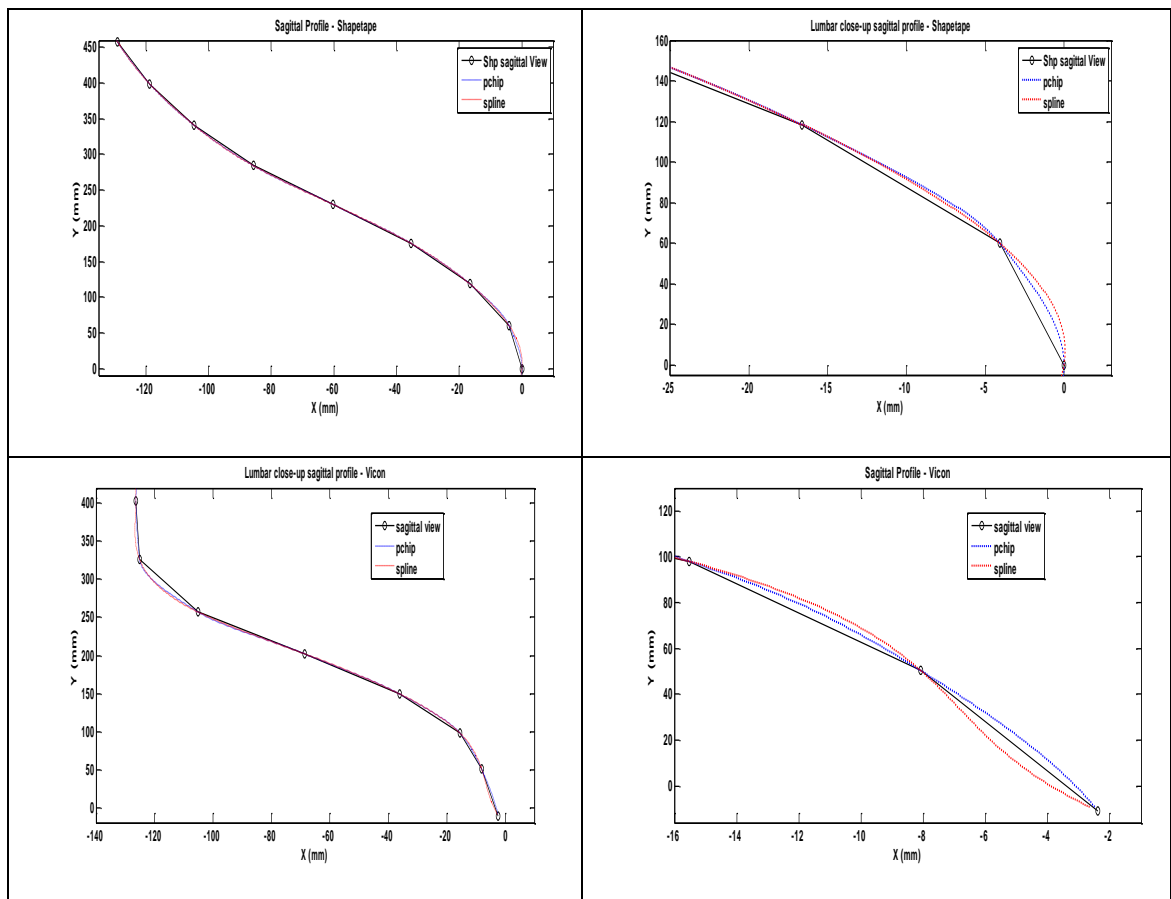


Figure A4.4. Spline fitting trials

Therefore all raw data were fitted with a piecewise cubic hermite interpolating polynomial (Matlab function pchip).

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